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Blood pressure waveforms

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BLOOD PRESSURE WAVEFORMS

by

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I. INTRODUCTION

Determination of the blood pressure waveform is of prime importance to both the medical practitioner and researcher. The methods of obtaining the blood pressure pulse fall into two general categories. First, the blood pressure pulse may be determined directly from the arterial system through intra-arterial means, this method is commonly referred to as the cannulation method. Secondly, the pressure pulse may be obtained through extra-arterial means. The purpose of this Thesis is to investigate the determination of blood pressure waveforms by extra-arterial means.

The historical background of the blood pressure pulse investigations will be reviewed. The hemodynamics of the blood system will be discussed and analysed as they apply to the extra-arterial determination of the pressure pulse. Various methods of determining the blood pressure pulse will be discussed.

Also to be presented will be a method of blood pressure waveform determination utilizing a resistive strain gage transducer and a fluid filled pickup in communication with the body surface. An analysis of the operation of the components of this system has been performed. The blood pressure pulse obtained by this system has been recorded for several conditions and the results of these recordings will be discussed.

II. BLOOD PRESSURE WAVEFORM DETERMINATION

A. Purpose of Blood Pressure Waveform Determination

The blood pressure waveform of the arterial system can provide significant information on the physical condition of the circulatory system. Wiggers (20) has shown that changes in the blood pressure waveforms from the established normal waveform can be attributed to disorders in the circulatory system, either in the heart or the vascular sub-system. Abnormal ventricular contractions in the heart will cause the pulse curve to deviate greatly from the normal pulse pressure representation.

Blood pressure waveforms will show the effect of changes in the vascular system. Of particular concern in this respect is the determination of hypertension. Goldring and Chasis (6) define the essential criterion for the diagnosis of hypertension as an abnormal elevation of the diastolic blood pressure. The diastolic blood pressure is the minimum pressure attained by the blood pressure waveform. Goldring and Chasis (6) show further that this rise in diastolic blood pressure is due to increased arterial resistance in the arterioles. With the rise in diastolic blood pressure there is a concomitant elevation in the systolic blood pressure, usually to the extent that the pulse pressure is increased. Systolic pressure is the maximum pressure attained in the arterial system. A rise in systolic pressure without a marked increase in the diastolic pressure will result from systolic hypertension which is different in pathological physiology and in the clinical course of treatment from diastolic-systolic hypertension (6).

Therefore, with the proper interpretation, the blood pressure waveform can be of considerable use to the medical practitioner in the diagnosis of disease. The usefulness of the blood pressure waveforms to the medical researcher will be limited only by the degree to which they represent the true "in vivo" waveforms.

B. History of Principles and Methods of Sphygmomanometry

The first actual measurement of blood pressure was performed by Stephan Hales in 1733. Hales used a direct approach consisting of inserting into an artery a brass tube to which was attached a glass tube. The level to which the blood rose in the tube was the measure of blood pressure. In 1828, Poiseuille introduced the U-form of mercury manometer which was used in place of the brass pipe of Hales. The disadvantage of the mercury manometer was that the effect of the large diameter of 7 mm combined with the heavy weight of the mercury column produced an instrument that was insensitive to small oscillations and pressure variations. The sensitivity was increased by the inclusion of a large reservoir and the reduction of the tube diameter by Mogendie in 1850.

In 1847, Ludwig introduced a method for direct registration of blood pressure which is still used in present investigations in essentially the original form. This scheme consisted of placing an ivory float in the longer arm of the Poiseuille manometer; a writing arm was attached to the float and the movements of the mercury column were transmitted to a writing surface.

The first indirect measurement of blood pressure in human beings was

made in 1834 by Julius Herrison. A hemisphere, the large surface of which was covered with a membrane, was connected to a graduated capillary tube and the entire system was filled with mercury. The membrane surface was placed on the radial artery and the oscillations of the column were observed. Vierordt, in 1855, introduced the principle of measuring the amount of counter-pressure to obliterate the pulsations in a peripheral artery, as an index of the blood pressure. Marey improved this technique in 1857 by using a water filled cylinder, into which the forearm was placed, as a means of applying pressure to occlude arterial flow. The fluid of the cylinder was in communication with a recording tambour. As the pressure of the fluid was increased the pulse amplitude increased at first and then diminished. Long before pulsations disappeared entirely the arm was seen to blanch, thus showing the vessels had been collapsed. Marey believed the point at which maximum oscillation occurred on the tambour was a true measure of the intravascular pressure. Later Janeway believed this point corresponded to diastolic pressure. Von Recklinghausen showed the external pressure which produced the maximum oscillation was actually closer to the mean blood pressure. The mean blood pressure is given (21, p. 22) as half the sum of the value of systolic and diastolic pressure. This is not strictly accurate as the mean value of a function P(t) over a time interval (a, b) is

$$M = \frac{\int_a^b P(t)\,dt}{b-a} \qquad 1$$

The value of M given by equation 1 will in general be nearer to diastolic

pressure than the value attained by dividing the difference between the systolic and diastolic pressure or pulse pressure by two.

From 1878 to 1896 several different models of sphygmomanometers were developed; however, all were based upon the principles of Marey. In 1896, Riva-Rocci devised an apparatus consisting of an arm cuff which encircled the middle of the upper arm. As the pressure in the cuff was increased the radial artery was palpated at the wrist. As the oscillations disappeared, the level of systolic pressure was read on the manometer. In 1897 an anaeroid manometer was attached to the cuff to measure pressure in place of the mercury manometer. The original arm cuff used was 5 cm wide and produced readings that were about 10 percent higher than expected. A cuff of 12 to 14 cm in width was introduced, which is still employed today, and gives accurate readings.

Korotkow first described the present technique used to determine blood pressure in 1905. This procedure is termed the auscultatory method and consists of using an arm cuff for compression of the radial artery, in combination with a stethoscope placed over the artery below the compressing cuff. Four distinct phases of sound are heard as the pressure in the occlusion cuff is decreased from occlusion. The sudden appearance of a clear faint tapping sound, growing gradually louder is termed phase I; phase II consists of the period during which the sound assumes a hissing, murmurish quality; phase III, the murmur disappears, and the sound becomes clear and louder; phase IV, the sound suddenly becomes muffled, and at the end of phase IV the sound has disappeared. Korotkow suggested that the systolic pressure be recorded at the beginning of phase I, and that the

diastolic pressure be recorded at the end of the fourth phase. Considerable disagreement over the correct interpretation of the fourth phase has resulted. Many investigators have found that the beginning of the fourth phase corresponds to the diastolic pressure found with the oscillometric methods when a sudden drop in maximal oscillation occurs. In 1939 the Committee on Standardization of Blood Pressure Readings of the American Heart Association recommended that this view be accepted as true measure of diastolic pressure. However, in 1951 the Council for High Blood Pressure Research of the Scientific Council of the American Heart Association concluded that the point of complete cessation of sounds is the best index of the diastolic pressure (10).

The current practice of using the point of sudden muffling of the sounds was based on comparisons with oscillatory criteria and on hemodynamic data derived from studies of isolated arteries. A practice based on oscillatory criteria alone seems of dubious value as no general agreement has been reached regarding the oscillatory criteria of diastolic pressure.

It should be noted that the present system of sphygmomanometry used depends on the auditory acuity of the investigator. A truly exact blood pressure measurement is not obtained, but an approximation (the value of which depends on the ability of the investigator) is made.

The use of intra-arterial catheters for blood pressure measurements provides a means of directly measuring the blood pressure. This method has the advantage of being capable of continuous recording or metering. With the introduction of proper corrections for instrument configuration,

the direct method of measurement can be expected to yield accurate results. The chief disadvantage of this method is the requirement of local surgical operative procedures due to the direct puncture of the arterial system.

III. CIRCULATORY SYSTEM

A. Functional Anatomy

The primary objective of the circulatory system is to transport materials to and from the tissues of the body (15). This involves the movement of many substances through the capillary walls into the cells of the tissues; it is dependent on the heart as a pump as well as the proper functioning of the arteries, arterioles, capillaries, veins, lungs, and numerous other organs and mechanisms of the body. The blood pressure waveform is created through the action of the heart pumping blood through the vessels of the circulatory system. The functional anatomy of the circulatory system will be reviewed with primary consideration given to the relation between the functions of the various parts of the circulatory system and the generation of the blood pressure waveform.

In general terms, it may be said that all the organs, bones, and tissues of the body are part of the circulatory system; however, the chief components of the system are the heart, arteries, arterioles, capillaries, venules, veins, and the circulation fluid, the blood. Figure 1 is a diagrammatic representation of the circulation system. The action of each component of this system in the development of the blood pressure waveform will be considered.

The function of the heart is to serve as a pressure pump to supply the energy required for the blood to flow through the circulatory system. The heart is essentially cardiac muscle divided into four chambers. The right and left atria serve as entrance chambers into which the blood flows

prior to entering the ventricles, which are the arterial pressure pumps of the circulatory system. The heart is divided by a septum into right and left halves. The two halves of the heart, each of which is composed of an atrium and ventricle, serve as two separate pumps operating in the same rhythm and phase. The right half of the heart serves to force the flow of blood through the lungs. The system composed of the right half of the heart and the lungs is the pulmonary circulation system. The system composed of the left half of the heart, the arteries, and veins is the systemic circulation system. These two sub-systems compose the circulatory system. Each ventricle has an entrance and an exit valve. The tricuspid valve is located between the right atrium and the right ventricle and is composed of three large flaps. The flaps are freely movable in the direction of inflow and are prevented from bulging and backflow during systole by the chordae tendinae and the papillary muscles. The bicuspid or mitral valve at the entrance of the left ventricle is similar to the tricuspid; however, it is composed of only two flaps. The exit valves from the ventricles are identical. These valves are composed of three cusps attached to the walls of the junction between the ventricle and artery. During systole of the ventricles the cusps are pressed against the walls of the aorta and pulmonary artery. At the beginning of the diastole or filling stroke the initial backward rush of blood into the ventricles from the arteries fills the cusps bringing them together to close the opening. It is of interest to note that there are no valves located at the entrance to the atria. Extensive backflow of blood is prevented by the contractural waveform of the atria which forces blood from

the venous input area toward the ventricles (7).

Contraction of the heart is initiated by the sino-atrial node or pacemaker. The pacemaker has the ability to initiate depolarization without external influence. Depolarization of myocardium causes the chambers of the heart to decrease in volume and thus force the blood into arteries. The pacemaker is governed by nerve impulses brought to it by the vagus nerve or the sympathetic nerves.

The flow of blood is thus seen to be from the venae cavae to the right atrium. Then through the tricuspid valve into the right ventricle from which blood is expelled at a systolic pressure of about 25 mm Hg (15). The blood then flows through the lungs and is oxygenated, and returns to the left atrium where it passes through the bicuspid valve into the left ventricle. This ventricle is in effect a pressure pump which pumps oxygenated blood into the aorta. Left ventricle systolic pressure is approximately 125 mm Hg in the adult human of 25 to 29 years of age (10). The blood exits from the left ventricle into the aorta.

The aorta is the chief vessel of the systemic system. The aorta distributes the blood through numerous branches to all parts of the body. The walls of the aorta and large systemic arteries are mainly composed of elastic fibers. The walls of the medium sized arteries contain more muscle tissue than the large systemic arteries. The function of this muscle tissue is to control the capacity of the circulatory system. The elasticity and contractural ability of the systemic arteries will be investigated further in a later section concerned with the hemodynamics of the circulatory system.

The arterioles are the terminal portions of the arteries and contain large amounts of circularly arranged smooth muscle. Through the contraction of the smooth muscle the inside diameter of the arterioles may be varied. Consequently, through consideration of the increased resistance to flow presented by the decreased diameter, the arterioles will greatly influence the rate of flow of blood from the arteries into the capillaries. This control of blood flow by the arterioles leads to the creation of an effective peripheral resistance (6). The smooth muscle of the arterioles is under a certain degree of continuous contraction or vasomotor tone. This tone is under the influence of the central nervous system.

Blood flows from the arterioles through the capillaries to the veins. The capillaries are approximately 8μ in diameter and are extremely thin walled. The average diameter of a red blood cell in man is approximately 7.5μ and the erythrocytes will therefore pass through the capillaries in a single column.

The blood is returned to the heart from the capillaries through the veins. The veins differ from the arteries in three chief aspects. First, the veins are thin walled and contain blood at low pressure in the order of 0 to 10 mm Hg while the arteries are relatively thick walled and have a mean pressure of 100 mm Hg (2). Secondly, the veins possess cusplike valves which permit flow toward the heart but hinder back flow, the arteries contain no valves. Thirdly, the veins are not elastic as are the arteries, but the veins do possess some muscle fibers which assist in the regulation of the capacity of the circulatory system to contain the

normal volume of blood.

B. Hemodynamical Considerations

The study of hemodynamics is concerned with movement of blood through the circulatory system. A systemic study of hemodynamics requires the consideration of several factors, some of which are heart rate, flow rate, blood volume, vessel elasticity, pulse pressure, effective peripheral resistance, and vasomotor tone. Factors such as blood flow rate or intra-arterial pressure are difficult to determine without at least a minor surgical operation.

Blood is a fluid with considerable viscosity. The dynamic viscosity of a fluid is given by the equation

$$\tau = \mu \frac{dv}{dy} \qquad (2)$$

where τ is the intensity of the fluid shear, and μ is the dynamic viscosity, and dv/dy is the velocity gradient with distance across the flow (13). Equation 2 is a statement of laminar flow. The velocity gradient may be interpreted as arising from adjacent fluid layers sliding over each other with different velocities. The velocity distribution of a viscous fluid flowing in a cylindrical tube as developed in the usual manner (13) is given as

$$v = -\frac{1}{4\mu}\left[\frac{D^2}{4} - r^2\right]\frac{dp}{ds} \qquad (3)$$

where dp/ds is the pressure gradient along the axial length, r is the radius of a cylindrical fluid element, symmetrical about the axis of the tube, and D is the diameter of the tube. It is seen that the velocity distribution over the tube cross section will be in the form of a paraboloid of revolution.

Blood is a non-homogenous fluid. It has been found that the flow of blood through small tubes, less than 1 mm in diameter, is altered from the laminar velocity distribution by two effects, the "sigma effect" and axial accumulation of cells (3). It was shown by Burton (3) that the integration for a differential laminae will not yield results agreeing with experimentally attained values. However, a summing process may be used to determine the value of effective viscosity in a tube of radius R with respect to the effective viscosity of a tube of infinite radius

$$4 \qquad \mu_{\text{RADIUS } R} = \frac{\mu_{\text{INFINITE RADIUS}}}{\left(1 + \frac{d}{R}\right)^2}$$

where d is the diameter of particles in the fluid. A value of viscosity will be attained that will agree with the experimentally derived value. Axial accumulation of red cells is explainable as a result of a Bernoulli force which tends to cause the cells to move towards the center of the flow stream. The net effect of axial accumulation is to reduce the effective viscosity of the blood. Burton (3) states that axial accumulation does not play a significant role in the physiologic range of blood flow velocities. Therefore, the effective viscosity is a constant, although this constant is less than it would be if there were no axial accumulation.

The Poiseuille-Hagen law is stated as the following formula for flow

(3)

5 $$Q = \frac{\Delta P \pi D^4}{128 \mu \ell}$$

where Q is the flow and ΔP is the pressure differential in the tube of diameter D and of length ℓ with fluid viscosity μ . From equation 5 we may consider a resistance to flow R_f which will be

6 $$R_f = \frac{\Delta P}{Q}$$

For Newtonian Fluids R_f will be a constant dependent on the values of ΔP and Q . However, for blood which is not Newtonian it was found that R_f does not remain a constant due to changes in μ and in D, the latter being the factor of greatest importance.

The relationship of blood flow and blood pressure has been seen to depend on the properties of the blood vessel walls. A consideration of the relationships between the force per unit area tending to deform a material or stress and the proportionate deformation of the material or strain, will aid in evaluating the properties of the blood vessels. As indicated in Figure 2, the stress acting in the direction of the circumference of the vessel wall is tangential tension, T. However, the measurable applied stress is the pressure which is applied in the radial direction.

To find the relationship between pressure and tension all forces under consideration are assumed to be in equilibrium. F_p will be the

force due to pressure and will include the effect of intravascular pressure. F_T will be the force due to tension. The force due to pressure will be equal to

$$F_p = PA_p \tag{7}$$

where P is the applied pressure and Ap is the area the pressure acts upon. Ap will be seen from Figure 2 to be equal to the product of the length $d\ell$ and the arc length $Rd\theta$. The force due to tension equals

$$F_T = TA_T \tag{8}$$

where T is the tension and A_T is the area the tension acts upon. The area acted upon is determined by the thickness S, the length $d\ell$, and the consideration that there are two surfaces upon which the tension acts.

Since all forces are assumed in equilibrium Fp may be equated to F_T providing proper correction is made to account for the different lines of action of these vectors. Therefore

$$F_p = F_T \sin \frac{d\theta}{2} \tag{9}$$

and upon simplification

$$T = \frac{PR}{S} \tag{10}$$

It is apparent that for a per unit of length of vessel the relationship between tension and pressure is both a function of vessel diameter and wall thickness.

Strain ζ , is defined as the change in dimension per unit dimension. Considering the vessel undergoing strain to be composed of a nonviscous, inertialess, linear, elastic material, Hook's Law may be applied and the tension is written as

11 $$T = E\zeta$$

where E is analogous to Young's modulus. The application of linear elastic considerations to the calculation of blood vessel mechanical properties is based upon the assumptions of small strain and of constant volume of wall material with strain.

Some studies (11) have shown that isolated strips of vessel have shown the wall to be visco-elastic, the strain being a function of the magnitude of the rate of change of stress as well as a function of the magnitude of stress. This effect may be written as

12 $$T(t) = E\zeta + K\frac{d\zeta}{dt}$$

where $T_{(t)}$ is a function of time and K is the strain-viscous coefficient. Since a real material must possess the property of mass, its effect may be added

13 $$T(t) = E\zeta + K\frac{d\zeta}{dt} + M_w\frac{d^2\zeta}{dt^2}$$

where M_w is the mass of the wall. From equation 10 the relationship of pressure to strain may be derived as

$$14 \qquad P(t) = E_p \xi + K_p \frac{d\xi}{dt} + \frac{M_w \delta}{R} \frac{d^2\xi}{dt^2}$$

Peterson et al (11) have shown that for a characteristic vessel where $\ell = 1$ unit; $\delta = 0.1$ cm; R = 0.5 cm and the density $\rho = 1$ gram/cm^3, the mass is 0.3 gram/unit length. Upon converting to c.g.s. units the equivalent mass coefficient becomes 0.0003 which is stated (11) to be negligible with respect to the values of Ep and Kp obtained. Therefore equation 14 may be written as

$$15 \qquad P(t) = E_p \xi + K_p \frac{d\xi}{dt}$$

The solution of equation 15 by classical means (17) yields a general solution of the form

$$16 \qquad \xi = e^{-\frac{E_p}{K_p} t} \int \frac{P(t)}{K_p} e^{\frac{E_p}{K_p} t} dt + C e^{-\frac{E_p}{K_p} t}$$

The exact form of equation 16 is dependent on the pressure function P(t) and the value of the constant C as determined by the conditions on ξ.

Peterson et al (11) have evaluated the pressure-strain elastic modulus Ep and the pressure-strain viscous modulus Kp for several animals and has demonstrated the effects of age upon the pressure-strain relationship.

It has been observed by many investigators (7, 10, 20, 21) that the

blood pressure waveform varies greatly as the pressure pulse traverses the arterial system. Figure 3 is a representation of the normal blood pressure waveforms as obtained by Wiggers (20). The top waveform was obtained from the carotid artery, the center waveform from the subclavian artery, and the lower waveform from the radial artery. While all three waveforms are of the same general shape it will be observed that considerable difference occurs in the systolic portion of each waveform and in the dicrotic notch which appears shortly after the systolic pressure has been reached. The dicrotic notch occurs in conjunction with the closing of the aortic valve. The effect of wave propagation may be observed from Figure 3. The subclavian artery is more proximal to the heart than either the carotid or radial arteries. The radial artery, for measurements made at the wrist, is more distal to the heart than either the subclavian or carotid arteries. The waveform appears at the subclavian artery measurement point prior to the time the waveform reaches either the carotid or radial measurement points. This effect will be investigated further in a following section. In this section the variation in shape of the pressure curve will be investigated.

Stacy and Giles (18) used a method of waveform matching to arrive at an equation which would represent the waveform variation as the pulse traverses the arteries. Figure 4 shows the model used in deriving the equation of waveform variation (18). The instantaneous upstream pressure waveform is given in terms of the instantaneous downstream pressure as

$$17 \qquad P_i = \frac{\rho L}{\gamma \pi R^2} \frac{d^2 P_o}{dt^2} + \left(\frac{\rho L}{2 \pi R^2} + \frac{8 \mu L}{\gamma \pi^2 R^4} \right) \frac{dP_o}{dt} + P_o$$

where μ is the viscosity and ρ is the density of incompressible fluid in the elastic tube of length L and radius R. The change in waveform of the pressure pulse is effected by changes in γ , R, μ , and Z. The quantities ρ , L, and π are fixed for the system being considered.

Equation 17 would be useful in computer simulation of waveform deformation in traversing the artery. The output pressure P_o could be operated upon using equation 17 and the result matched to P_i. This procedure would enable the attainment of coefficients for the equation. Analysis of this type would aid to a more thorough understanding of the propagation of the blood pressure pulse and the actions of the circulatory system.

C. Electrical Analog of the Arterial System

In the field of Electrical Engineering considerable effort has been spent on the analysis of wave propagation on transmission lines. The transmission line may be considered as an analog of the circulatory system. Figure 5 indicates an ideal transmission line, or a line which does not possess components capable of energy dissipation. The arterial system, which in the circulatory system corresponds to a transmission line, does possess energy dissipating elements such as the muscle of the vessel wall and the viscosity of laminar flow. However, for the purpose of developing a simple analog to the circulatory system the arterial system will be considered to be lossless and μ will be considered as zero. It should be noted that the venous side of the circulatory system is considered as possessing zero pressure.

Aseltine (1) states the requirement for an analog to be that the equations for each system be of the same form. The fundamental equations (12)for the analysis of a transmission line as shown in Figure 5a are

18 $$\frac{\partial V}{\partial z} = -L \frac{\partial I}{\partial t}$$

19 $$\frac{\partial I}{\partial z} = -C \frac{\partial V}{\partial t}$$

These equations may be combined into an equation containing voltage alone or current alone. An equation containing voltage only is

20 $$\frac{\partial^2 V}{\partial z^2} = LC \frac{\partial^2 V}{\partial t^2}$$

An equation containing current only is

21 $$\frac{\partial^2 I}{\partial z^2} = LC \frac{\partial^2 I}{\partial t^2}$$

Peterson et al (11) have shown that a relationship between pressure and volume changes may be written as

22 $$\Delta P = \frac{\rho L}{\pi R^2} \frac{d^2 V}{dt^2} + \frac{8 \mu L}{\pi^2 R^4} \frac{dV}{dt}$$

Using the value of μ as zero, the equation relating pressure and volume in an artery may be written as

23 $$\frac{\partial P}{\partial z} = -\frac{\rho}{\pi R^2} \frac{\partial^2 V}{\partial t^2}$$

Relating the flow, Q, to volume changes by $Q = dV/dt$;

$$24 \qquad \frac{\partial P}{\partial z} = -\frac{\rho}{\pi R^2} \frac{\partial Q}{\partial t}$$

The rate of change of flow with distance can be considered as the product of an elastic capacity factor C' and the rate of change of pressure with time.

$$25 \qquad \frac{\partial Q}{\partial z} = - C' \frac{\partial P}{\partial t}$$

The capacity factor relates the change in volume to a change in pressure.

Combining equations 24 and 25 into one equation containing the pressure only

$$26 \qquad \frac{\partial^2 P}{\partial z^2} = \frac{\rho C'}{\pi R^2} \frac{\partial^2 P}{\partial t^2}$$

An equation containing only flow may also be obtained

$$27 \qquad \frac{\partial^2 Q}{\partial z^2} = \frac{\rho C'}{\pi R^2} \frac{\partial^2 Q}{\partial t^2}$$

Equations 26 and 27 are of the same form as equations 20 and 21. Therefore the transmission line may be considered an analog of the arterial system. Table 1 shows the correspondence between quantities normally used in the analysis of arterial systems and those used in the analog.

Equation 26 is a differential equation of a propagating waveform

whose solution may be found by classical means (5) as

$$28 \qquad P(t,z) = G_1\left(t - \frac{z}{v}\right) + G_2\left(t + \frac{z}{v}\right)$$

where the velocity of wave propagation

$$29 \qquad v = \sqrt{\frac{\pi R^2}{\rho C'}}$$

Equation 28 indicates that the pressure may be considered as composed of a wave traveling in the positive z direction with velocity v and a wave traveling in the negative z direction with velocity v.

The velocity of propagation may be computed by equation 29. C' is determined from consideration of the change in flow for a differential segment of artery dz. The net change in flow for any instant will be

dz and is merely that flow that is absorbed in the change in the volume of the vessel. From the cardiac ejection volume V_E, a net change in flow for one heart beat may be calculated. Assuming that for a 70 Kgm normal adult with a heart rate of 72 beats per minute the mean heart output is 5 l. p. min. (19, p. 720). V_L may be calculated as 69.5 ml or a net flow change of 69.5 ml p. min. Assuming a systolic pressure of 130 mm Hg and a diastolic pressure of 80 mm Hg, the elastic capacity is

$$30 \qquad C' = \frac{69.5 \times 10^{-2}}{(130-80)(13.59)(980)} = 1.05 \times 10^{-3} \ cm^3/dyne/cm^2$$

The factor $\rho/\pi R^2$ is related to the mass of the fluid. For the example

under consideration the πR^2 term may be considered in conjunction with the effective arterial length to give a $\pi R^2 L$ term which is the volume factor V. Assuming 19 percent of the volume of blood is in the arterial system, a volume factor V of 950 ml is obtained. The density of blood may be assumed to be 1.03 gm/ml (3). The velocity of propagation is then 9.4 m. p. sec. The measured propagation velocity is 9 m. p. sec. in normal man (2). Therefore, the close agreement between the measured and calculated values give substance to the use of a transmission line for an analog of the arterial system.

D. Extra-arterial Blood Pressure Waveform Determination

A device to obtain the blood pressure waveform of the circulatory system by extra-arterial means must conform to several considerations based on the anatomy and hemodynamics of the body (8). The anatomical structure of the body is such that the large arteries are positioned in the medial part of the body and pass close to the body surface at only a few isolated points at which the pulse may be palpated. The aorta is medial in the body cavity and branches into the femoral arteries which continue to the legs. The femoral artery is located close to the body surface on the anterior side near the junction of the leg with the lumbar section of the body. The carotid artery is located lateral to the trachea and is near the surface on a small area of the neck. The temporal artery appears particularly close to the surface in the area in front of the ear lobe and anterior to the zygomatic arch. This area is easily located by the strong pulse palpated there. Strong pulses may be palpated in the

frontal branch of the temporal artery at the area superior and posterior to the eye and in the angular artery, lateral to the bridge of the nose.

Other locations where arteries are particularly proximal to the body surface are the area where the subclavian artery passes beneath the clavicle; the wrist area where the radial artery passes proximal to the surface; and on the portions of the fingers and toes where the digital arteries are proximal to the surface.

The areas where the pulse may be palpated as described above have two chief points in common. First, these areas are locations where the artery passes over a relatively hard surface, such as bone or the cartilage of the trachea. Secondly, these locations are marked by the absence of muscle tissue between the artery and the skin. These locations may then be characterized as having an artery surrounded by fascia passing between a bone structure and the skin.

The temporal artery anterior to the ear lobe and the digital artery of the finger were selected as the chief artery locations to be used in obtaining the blood pressure waveform. These locations will be termed arterial pulse areas.

IV. ACQUISITION OF BLOOD PRESSURE WAVEFORMS

A. Systems of Obtaining Blood Pressure Waveforms

The primary purpose of this Thesis is to obtain blood pressure waveforms from the human by extra-arterial means. Several secondary objectives are also to be considered. First, the device for obtaining the waveform must reproduce a representation of the waveform actually in existence within the artery. Second, the device used to obtain the waveform must be small, easy to operate and be reliable. Third, the device should be capable of continuous blood pressure waveform recording for extended periods of time, in excess of an hour. During extended periods of recording the subject should be permitted to function in a normal manner without being hindered or affected by the device. Fourth, the device should be able to produce an output signal which is free from the effects of muscle movement. The last consideration is that the system to obtain the waveform should utilize the capabilities of existing equipment wherever possible.

The major principle to be followed in obtaining blood pressure waveforms is the use of an applied balance pressure to occlude the artery and a membrane placed against the body surface to obtain the actual waveform. Historically, systolic and diastolic pressures have been determined using an adaption of this principle. In a previous section the anatomical considerations of obtaining the waveform were presented. The application of the basic principle described above to the previously mentioned arterial pulse areas will be studied. The "floating form" of the blood pressure

waveform may be obtained from a membrane placed against the skin. The term "floating representation" is used to mean the pulse pressure waveform without reference to the actual numerical values of pressure of an intra-arterial blood pressure waveform. The pulse waveform is transmitted to the membrane which is in communication to a device for pressure to an electrical signal conversion, or a transducer. The transducer output may be amplified and then applied as an input to an oscilloscope or oscillograph. A "floating representation" of the blood pressure waveform is capable of being recorded either photographically or by recorder tracing.

While it is desirable to obtain ordinate pressure values for the entire blood pressure waveform, extra-arterial measurements will not provide true pressure readings from the calibration of the transducer pickup unit itself. The actual blood pressure of the artery under investigation is considerably higher than any pressures measured at the surface of the skin. It is necessary to actually calibrate the pressure system composed of the artery under investigation, the pickup and the recording device. The introduction of a standard pressure waveform into the arterial system enabled this system to be calibrated and the relative coefficient between the actual pulse pressure and the represented pulse pressure was established. A standard pressure cannot actually be introduced into the artery. However, a standard pulse calibration may be effected by applying a pressure to the arterial pulse location, which will occlude flow in the artery. The standard pressure is then a pressure which will just balance the pressure in the artery; and thus cause the pressure waveform in the artery to be reduced to zero. The pressure which balances the maximum

pressure of the arterial system is systolic pressure. The pressure waveform may be monitored through the transducer-recorder system at a location downstream to the area of external pressure application. It should be noted that the application of this principle to a system containing a waveform varying at a low frequency rate, approximately 14 cycles per minute in the human, about some mean pressure value in addition to the higher frequency variation of the waveform (approximately 72 cycles per minute in the human) will result in an approximation to the actual instantaneous systolic pressure. It is generally accepted however, that the approximation made to systolic pressure is a good measure of the actual instantaneous systolic pressure in the artery. Goldring and Chasis (6) reported values of systolic pressure measured by the auscultatory technique which average 10 mm Hg lower than values obtained by direct arterial measurement. If the pressure applied to the arterial pulse location is decreased from systolic pressure the pulse amplitude will increase. When the pressure in the artery is equal to the externally applied pressure, the artery will resume a normal cross-section configuration. Therefore, the pulse waveform obtained at the monitor location will be of a maximum amplitude. A measure of diastolic pressure is the pressure applied externally that will just produce a maximum amplitude in the pulse waveform.

Through determination of the systolic and diastolic pressures of the blood pressure waveform, the maximum and minimum values of the "floating waveform" or pulse waveform have been obtained. If the arterial transduction and recording system is linear in the reproduction of the pressure waveform representation, pressures intermediate to the systolic and dias-

tolic pressure may be obtained through linear interpolation of the recorded output curve.

The method of calibration of the pressure waveform as described above has one chief disadvantage. The pressure waveform curve cannot be obtained during the determination of systolic pressure. The entire pulse waveform is not obtained until the external pressure is less than diastolic pressure of the blood. Furthermore, the actual process of obtaining the systolic and diastolic pressure values requires several pulse cycles. Therefore, it is desirable to calibrate only as often as required by variations in the actual pressure waveform being measured.

Some of the possible design configurations which meet the above requirements are shown in Figure 6. Figure 6a is a block diagram of a device which is placed against the arterial pulse area with the pickup diaphragm downstream with respect to the occlusion plunger, which is positioned to apply an occlusion force to the artery. A solenoid coil, whose coil current is proportional to the applied force, could serve as the occlusion motive unit. The coil current may be used as a measurement of the applied force. The pulse waveform is transmitted from the pickup diaphragm to the transducer, through the connecting fluid of the connecting chamber. The transducer output, $P_1(t)$, is a representation of the pressure function impressed on the diaphragm. Figure 6b is a block diagram of a device similar to the one described above, except a pressure source and a transducer are substituted for the solenoid motive unit. The representation of the applied pressure, $P_2(t)$ is the output signal of transducer 2.

The requirement of a small, simple device suggests the use of only one transducer. Figure 6c is a block diagram of a device which requires only one transducer. The applied pressure which is transmitted to transducer 2 in Figure 6b is now transmitted to a connecting chamber and then to the single transducer.

Figure 7 is a representation of the output waveform of the device of Figure 6c. At time A the pressure in the device is increased from the pressure required to obtain pulse pickup. As the external pressure is increased the pulse waveform decreases in a manner described above. When the pulse waveform $P_1(t)$ has been reduced to zero by the externally applied pressure $P_2(t)$ for an interval of time equal to the average period of the pulse waveform, the applied pressure is reduced. As the pressure is reduced to the pickup pressure the pulse waveform will reappear at time C and a pressure corresponding to the systolic pressure is noted. As the externally applied pressure is further reduced the pulse waveform will reach a maximum amplitude at time D. The pressure at this time will correspond to diastolic pressure. The external pressure is reduced to the pickup pressure at time E.

It should be observed that the accuracy of the systolic and diastolic pressure determination will be effected by the rate of change of the external pressure $P_2(t)$. An externally applied pressure which more fully utilizes the characteristics of the arterial system is shown in Figure 8. The pressure is increased from the pickup level at time A to $P_2(B)$ at time B. The value of $P_2(B)$ is established as a variable limit constant of the system. The pressure is reduced at a predetermined rate to $P_2(C)$ which is

systolic pressure and corresponds with the appearance of the first cycle of the pulse waveform. The pressure is then rapidly reduced to $P_2(D)$, another variable limit level of the system, after which the pressure is reduced at a slow fixed rate until the maximum amplitude of pulse waveform is attained. $P_2(E)$ corresponds to diastolic pressure and at time E the externally applied pressure is reduced to the pickup pressure level at a rapid rate.

The filter shown in Figure 6c is necessary to remove the pulse waveform that is transmitted to the transducer even though the applied pressure exceeds the systolic pressure of the artery. A diagram of the artery and occlusion diaphragm are shown in Figure 9. The pressures in the occlusion diaphragm cause the artery to be occluded, however, a portion of the artery in contact with the diaphragm will still receive the blood pressure waveform. This section of the artery is marked (A) in Figure 9. The portion of the pressure waveform transmitted through the skin and diaphragm from section A could be interpreted erroneously as failure of the pressure diaphragm to have occluded the artery. The filter, constructed of a small bore tube (.58 mm inside diameter) and a distensible chamber, has been experimentally evaluated and the results of this evaluation will be discussed in a following section.

An electrical circuit scheme which will perform the operation described in Figure 7 with the applied pressure locus of Figure 8 is shown in Figure 10. The output of the transducer of Figure 6c is passed to a clamper and a smoothing circuit. The pulse waveform as actually recorded is considerably more reduced in amplitude with respect to the $P_2(t)$ curve

than is shown in Figure 7. The clamper will remove the effect of the applied pressure. The clamped pulse waveform will be increased in amplitude by the amplifier. The output of the amplifier is available for direct recording of the pulse waveform, or "floating waveform". The amplifier output is an input to a differentiation circuit. One section of the differentiation circuit will differentiate the pulse waveform, but permit only the differentiated waveform of the first pulse after occlusion to pass to the pulse shaper. The other section of the differentiator will obtain the envelope of the pulse waveform and differentiate this envelope and provide an output pulse to the pulse shaper when the rate of change of the envelope of the pulse waveform immediately following occlusion becomes zero. The pulse shaper will shape the two pulse inputs into rectangular pulses of short duration.

The output of the pulse shaper is combined with the output of the smoothing circuit, which is the same as Figure 8, and provides a waveform to the recorder as shown in Figure 11. The output from the smoothing circuit is fed to a comparison network which compares this waveform with the values of pressure established in the equilibrium limit control unit as explained in connection with Figure 8. The output of the comparison control unit is in the form of a control signal to the pressure source unit. A sequence timer provides an input to the pressure sources and serves to initiate the pressure control system. In the absence of the control inputs to the pressure source, the amplifier provides a waveform to the recorder similar to Figure 7, prior to the time A.

The devices presented above require a means of conversion of the

pressure signal in the connecting chamber to an electrical output signal. The selection of a suitable transducer will be discussed in the following section.

B. Transducer Selection

A review of the various methods of pressure to electrical conversion revealed four general types of transducers. These transducers were resistive strain gages, piezoelectric crystals, variable capacitance devices, and differential transformer devices. A fifth scheme for transduction, consisting of an oscillating mass whose frequency of oscillation is proportional to the mean pressure of an elastic vessel was derived and is presented in the appendix.

In consideration of the requirements for a small transducer component, requiring a relatively small amount of associated circuitry and capable of providing undistorted transduction, the resistive strain gage was selected as being preferable over the other methods of transduction. An evaluation of several resistive transducers was made. A resistive strain gage transducer capable of representing pressures to 50 mm Hg was located in the Veterinary Physiology Department at Iowa State University. This transducer will be identified by manufacturers type number which is P23B. This transducer will enable the recording of extra-arterial pressure waveforms but will not suffice to transduce the occlusion pressure.

A miniature resistive strain gage transducer, available from Statham Laboratories, Inc., was selected because of the values of pressure range,

natural frequency response, and output sensitivity. The dimensions and specifications of this transducer PG222 are shown in Table 2. This transducer is to be used in conjunction with a Tektronix Type 502 Oscilloscope having a direct coupled amplifier with sensitivity to 200 μ v/cm deflection.

A device to record the pressure waveform should possess a flat frequency response over as wide a frequency range as technically feasible. Since it is planned to use a fluid medium to match the transducer to the arterial pulse area a damping effect due to the liquid flow will be incurred. Therefore, a transducer with a high undamped natural frequency was chosen in order to obtain an essentially flat output response over a wide frequency range.

The devices discussed above and shown in Figure 6 require two separate transducer systems. The characteristics of the two transducers used are shown in Table 2. Figure 12 is a block diagram of the basic transducer system used. The pressure input to the transducer is converted to an electrical signal and displayed on a recording device. Excitation power is provided from the power unit. A zero reference level and a calibration signal are provided by the zero balance and calibration controls respectively. Figure 13 is a schematic diagram of the power supply, zero balance control, and calibration control. Each transducer is essentially an unbounded strain gage connected in the configuration of a Wheatstone Resistance Bridge. The bridge circuit for the PG222 transducer also includes two temperature compensating resistors which are in series with the input leads. An external precision variable resistor will be used as a

calibration control unit. This calibration resistor is shown as Rc and is only connected into the circuit when the key switch is depressed. The valve of Rc is calculated to reproduce an output representative of a given pressure applied on the transducer diaphragm. The transducer manufacturer, Statham Laboratories, Inc., suggests the following equation

$$R_c = \frac{10^6 E R_o}{4NF} - \frac{R_{in} + R_o}{4} \qquad (32)$$

where the values of E, R_o, R_{in}, and F are given in Table 2. The quantity N is the number of mm Hg acting upon the transducer diaphragm represented by the calibration resistor. The calibration resistors for a representative deflection of 100 mm Hg for the PG222 transducer and 20 mm Hg for the P23B transducer are given in Table 2. Two Power-Balance units have been constructed. These units were mounted on a test board with facilities for connecting a mercury manometer into a test system in parallel with the pressure transducers and waveform pickup devices. A hand pressure bulb was incorporated on this test board as a source of pressure. Figure 14 shows this test board. The two power-balance units are shown on the rear section of the test board. The battery power supplies used for each transducer are external to this board and are not shown in Figure 14.

The P23B transducer in the center foreground is connected to an adjustable arm rest. The PG222 transducer is shown placed on the arm rest pad. Figure 15 shows both the P23B and PG222 transducers. The scale marking in Figure 15 is inches. The larger transducer is the P23B transducer. The valves on the chamber end of the P23B transducer are to facili-

tate rapid connection of test apparatus to the transducer. The PG222 transducer is shown with an adapter to facilitate the connection of the transducer to the system to be evaluated. This adapter has the effect of increasing the transducer size, but in the interest of interchangeability and transducer protection the adapter will be used in all evaluations. The axial diaphragm displacement for the PG222 transducer is 3 by 10^{-3} inches for full-scale deflection. This deflection represents a volume change of 14.73 by 10^{-5} cubic inches. This volume change is accompanied by flow through the .05 inch diameter bore of the adapter. The pressure differential introduced by flow through the adapter is negligible for the frequency range of investigation.

The transducer output signal is applied as an input to a recording device. Two recording systems are used. First, an oscilloscope-camera recording system consisting of a Tektronix 502 High Gain Oscilloscope and a Polaroid Scope Camera which is connected to the transducer output through the power-balancing unit. The Tektronix 502 Oscilloscope was selected as a component of the recording system because of the low level DC amplifier contained in this unit. The amplifier is stable and has inherent open terminal noise of less than 50 μ v. A further attraction of this oscilloscope is the availability of a dual trace display through separate vertical amplifiers. This feature permits the simultaneous comparison of waveforms. The oscilloscope-camera recording system is utilized in the experimental evaluation of systems and devices to obtain the blood pressure waveforms.

Another recording system utilized in the experimental evaluation pro-

cedure is a Sanborn Model 60 Recording Cart. This recorder has a low level preamplifier which permits the display of signals with a sensitivity of approximately 2.5 mv/cm deflection. This recorder is also a dual trace device and permits the simultaneous recording of two independent waveforms. Figure 16 shows the Tektronix Oscilloscope and the Sanborn Recorder. Warner (19) has shown by means of Graphical Fourier Analysis of the blood pressure waveform obtained intra-arterially that the 10th harmonic of the blood pressure waveform is less than three percent of the amplitude of the fundamental. The fundamental frequency of the normal blood pressure waveform of the human is 1.2 cps. Therefore, the 10th harmonic is approximately 12 cps. The establishment of a value of maximum frequency response of 60 cps or approximately the 50th harmonic should provide a recording system capable of reproducing the essential information of the blood pressure waveform as applied to the diaphragm of the transducer. The frequency response of PG222 transducer, the power balance unit, and the oscilloscope recorder is essentially undistorted. The Sanborn recorder has an output response that is flat to within 3 db for frequencies from 0 to 45 cps. The recording system composed of the transducer and Sanborn Recorder has a maximum useful frequency of 45 cps. This system was used to obtain system tracings similar to Figure 7. The maximum frequency of 45 cps will suffice for this application. The frequency response characteristics of the P23B transducer are essentially identical with the system characteristics presented above.

One further component is used in the evaluation of the blood pressure measurement systems. This component is a low level Tektronix 122 Physio-

logical Preamplifier with a gain of 1000. This preamplifier is capacitive coupled and is shown on the right of Figure 16.

V. EXPERIMENTAL PROCEDURES

A. Arterial Pulse Area Evaluation

An experimental evaluation of the systems described in the previous sections has been conducted. The concept of an arterial pulse area was investigated and the area most feasible for obtaining the blood pressure waveform has been determined. Devices for obtaining the blood pressure waveform have been constructed and evaluated.

The first area of experimental investigation was the evaluation of the arterial pulse areas as described in part III section D. The arterial pulse areas were studied to determine the relative strengths of the pulse waveform at the skin surface and the determination of methods of applying external occlusion pressure.

The temporal arterial pulse area was investigated first. The pulse waveform was obtained by utilizing the P23B transducer. A rubber diaphragm held in a cylindrical diaphragm holder was placed against the skin over the temporal artery. The transmission of pulse waveform through various membrane configurations was evaluated. The diaphragm holder and the diaphragm configurations evaluated are presented in Figure 17. Rubber of approximately .1 mm thickness was used to form the diaphragm. The anatomical structure of the arterial pulse area as shown in Figure 9 suggested the use of a diaphragm as shown in Figures 17b or 17d.

A test platform, as shown in Figure 18 was constructed. The cap and associated equipment are worn on the head. The sliding diaphragm holder jig is designed to be placed over the left temporal arterial pulse area.

The diaphragms shown in Figure 17 were evaluated by the following procedure. First, a diaphragm was placed in the holder. The holder was connected to the P23B transducer by a 150 mm length of 7 mm inside diameter polyethylene tube. The tube has a wall thickness of 2 mm and exhibits no apparent deformation with pressure over the pressure range of the transducer. The holder and connecting tube were then placed into the jig. The cap assembly was placed on the head and the sliding jig positioned over the temporal area. The jig was adjusted against the skin by means of a 30 thread per inch screw until a pulse waveform was obtained. A system of mirrors was constructed to permit self application of this jig. The determination of pulse waveform was made by displaying the transducer output on the oscilloscope as described in the previous section. The jig was adjusted to obtain an output waveform representation of maximum amplitude.

Waveforms obtained with the oscilloscope camera recording system for the experimental procedure described above are shown in Figure 19. The waveforms displayed are the waveforms of maximum amplitude. The associated values of pressure and output voltage are summarized in Table 3. An evaluation of the quality of waveform representation is difficult due to the lack of data for blood pressure waveforms obtained from the temporal artery either extra or intra-arterially. It will be noted the waveform recorded from the arterial pressure pulse area is only 6.1 mm Hg for the greatest waveform output.

Two other important factors were evaluated at the temporal pulse area. Variations in the amplitude of pulse waveform with changes in pick-up location were studied. It was found that a variation along a surface

path which was perpendicular to the artery resulted in a reduction of pulse waveform amplitude by almost 90 percent for a 1.5 cm change from the point of maximum pickup. The effect of muscle movement on the output of the transducer was pronounced. The masseter muscle is located proximal and caudad to the temporal arterial pulse area. Consequently, as this muscle was exercised there was a resultant skin movement which was transmitted to the transducer and produced a distorted output signal. The cap, platform, and transducer added to the effect of muscle movements. Their relatively large mass had the effect of creating false signals as the head was moved. Methods of applying an occluding pressure to the temporal arterial pulse area were studied.

An overall evaluation of the temporal arterial pulse area indicated that this area is not suitable for extra-arterial pulse measurements to be conducted on a normal, unrestricted subject using a miniature pickup transducer system.

The angular arterial pulse area was investigated to determine the feasibility of utilizing this artery for blood pressure waveform measurements. A jig similar to an eyeglass frame was constructed. This jig is used in conjunction with the cap described previously.

The spheroidal membrane of Figure 17b was tested at this area. A maximum transducer output of approximately 100 μ v was obtained. This relatively low value of output voltage as compared with the output voltages of the temporal area and the possible injury to the eyes in the event of a blow against the eyeglass jig prompted the discontinuance of investigation on the angular arterial pulse area.

The digital arterial pulse area was investigated next. Specifically, the medial artery of the index finger of the left hand was studied. Consideration of the requirement for a small pickup device prompted design of a pickup diaphragm which would be more suited for finger application than the cylindrical holder used previously. Several finger pickup designs were considered. Figure 20 shows four of the basic finger pickup diaphragms that were constructed and tested. The pickup shown in Figure 20a was used to perform initial studies of the digital arterial area. This pickup was placed on the proximal section of the finger. It was observed that if the diaphragm was wrapped tightly around the finger and the transducer diaphragm system was filled with distilled water to a pressure of 30 mm Hg, an output waveform similar to that of Figure 21 was obtained. The pickup was wrapped around the finger so as to orient the luer connector on the backhand side of the finger. The pickup was connected to the P23B transducer as shown in Figure 22. Care was taken to remove all air bubbles from the fluid connecting system. The transducer was connected to the oscilloscope as described previously. Figure 21 shows the "floating representation" of the blood pressure waveform obtained from the digital artery. The maximum pulse waveform obtained with this pickup transducer system was approximately 2.5 mv. Figure 23 shows a comparison of the waveforms obtained from the temporal and digital pulse areas. Figure 23a was obtained with the spherical pickup and Figure 23b was obtained with the pickup shown in Figure 20a.

The anatomical structure of the digital arterial area permits the application of an external occlusion pressure with an occluding bag bound

around the pulse area.

B. Waveform Determination from the Digital Pulse Area

The initial investigations of the digital pulse area prompted further consideration of the digital pulse area as an operating location from which blood pressure waveforms could be obtained.

The standard experimental procedure was to place the left forearm of the subject on the arm rest, shown in Figure 14, and attach the device to obtain the blood pressure waveform to the index finger. A 15 cm standard occlusion bag was attached to the left upper arm in the normal clinical manner. Air pressure was supplied to the bag from the pressure system of the test board. Provision was made to monitor the application of pressure by attaching the PG222 transducer to the pressure system. As the pressure in the occlusion bag was decreased systolic and diastolic pressures were determined by the auscultatory method as described by Korotkow.

The test board was used in determining the relative waveform transmission of the pickups shown in Figure 20. The pulse waveforms as transmitted from the digital pulse area through the pickups shown in Figure 20a, b, c and shown in Figure 24a, b, c respectively. It will be seen that the pickup of Figure 20a exhibits properties of "ringing". Through repeated waveform determination, the pickup shown in Figure 20b was found to reproduce a waveshape that is essentially the same form as waveshapes presented by Wiggers (21).

Utilizing the pickup shown in Figure 20b as a standard pickup, investigations were conducted to determine the most satisfactory method for

obtaining the values of pressure which correspond to the ordinate values of the "floating representation" of the blood pressure waveform. The procedure used was to place an occlusion cuff around the proximal phalanx and the pickup around the middle phalanx. Both the pickup and occlusion cuffs were wrapped securely with a nonstretchable nylon bandage and fastened with standard medical adhesive tape. This procedure will permit normal flow in the digital artery until the pressure in the occlusion cuff is increased to effect occlusion. This method of attaching the occlusions cuff and pickup is shown in Figure 25.

The first step in the investigation of arterial occlusion with the finger cuff was to establish a procedure by which systolic and diastolic pressure could be obtained from the "floating representation" of the pulse waveform in combination with the externally applied occlusion pressure. Systolic and diastolic pressures for several subjects were taken by the auscultatory method and recorded. The pressures in the occluding cuff of the upper arm were recorded in conjunction with the pressure pulse waveform. An example of this recording is shown in Figure 26. The time markings represent seconds. The pressure pulse recording is the lower trace. The occlusion pressure is shown in the top recording trace. The erratic variations in this trace were due to noise from the recorder preamplifier.

At a later time, without consultation of the values of systolic and diastolic pressure obtained by the auscultatory method, systolic and diastolic pressure of the recorded pulse waveform were read from the recordings. Systolic pressure was recorded as the pressure in the oc-

clusion cuff at the instant the first pulse of the pulse waveform appeared. Diastolic pressure was recorded as the pressure corresponding to the instant the pulse waveform reached a maximum peak to peak amplitude. The pressure at which the dicrotic notch first appeared in the waveforms was also recorded. Table 4 is a compilation of the data of several tests. Over 30 pressure calibration runs have been performed. Table 4 shows systolic pressure determined from the waveform recordings to average 1.3 percent lower than values obtained by the auscultatory method. The variation in diastolic pressure is 5.9 percent lower for values obtained from the waveform recordings.

It should be noted that the values of systolic and diastolic pressure are for a particular instant of time. The determination of the systolic pressure by both the auscultatory and waveform methods provides values of systolic pressure that are in close agreement. A possible explanation for the difference between the values of systolic pressure rests in the observation that the mean blood pressure varies in direct relationship with the breathing rate. This variation in mean pressure causes a variation in the pulse waveform representation. This breathing effect on the mean pressure has been observed on several recordings. The average period of the breathing response is approximately 4.5 seconds. Since the time required to obtain a waveform representation such as Figure 26 is in excess of 5 seconds, the effects of respiratory pressure changes will be present on the waveform. Figure 27 shows the variation in digital pulse waveform with respiration. If the occlusion pressure is rapidly decreased from systolic pressure, the instant at which the pulse waveform actually at-

tains a maximum value is difficult to decern using the recorded curve. An erratic decrease in pressure will also produce a waveform that is not useable in the determination of diastolic pressure. A pressure release valve is incorporated in the pressure source mechanism to insure a smooth decrease in pressure. A combination of waveform distortion through the variation of mean pressure due to respiratory effects and to the erratic decrease of the occlusion pressure will explain the differences between measurements by the auscultatory and waveform methods. If the maximum amplitude of the waveform is to be taken as a measure of the attainment of diastolic pressure, the effects of the mean pressure changes must be corrected. The data in Table 4 has not been corrected for mean pressure changes.

The schematic diagram shown in Figure 10 incorporated a circuit which will correct for mean pressure changes. This circuit will produce a marker pulse corresponding to the time at which the pulse waveform reaches maximum amplitude. This marker pulse can be correlated with the occlusion pressure recording to obtain diastolic pressure.

While the method of calibrating the pickup pulse waveform representation as described above does not actually compare pickup pressures with true intra-arterial pressures, this method does compare the pickup data with values of pressure as obtained in normal clinical practice. An actual intra-arterial pressure correlation would require a cannulation of the artery simultaneously with the pickup measurement. The facilities needed to perform this correlation calibration on a human are not available at Iowa State University. One of the original specifications for a device to

obtain waveforms was that the device be useable to the medical practitioner; therefore, a calibration in terms of measurements that are familiar to the practitioner and that are tabulated as diagnostic aids (10), should ultimately be the most satisfactory calibration.

Utilizing the criterion for systolic and diastolic pressure determination developed above, the pressure in the occlusion cuff around the proximal phalanx was varied from above the anticipated occlusion pressure to zero pressure. The occlusion cuffs tested are shown in Figure 28. The narrow cuff of Figure 28 has a deflated width of 19 mm and the wide cuff has a width that varies from 15 mm to 40 mm. It was found that pressure measurements made with the narrow occlusion cuff were approximately 30 mm Hg higher than measurements made by the auscultatory method. Measurements made with the wide pressure cuff were approximately 25 mm Hg lower than auscultatory values. An occlusion cuff width of 25 mm, measured directly over the digital artery, produced systolic and diastolic pressures that corresponded favorably with auscultatory measurements. The results of these measurements are shown in Table 5. A recorded waveform for this measurement comparison test is shown in Figure 29. It was observed that all measurements made using a finger occlusion cuff approximately 25 mm wide yielded values of systolic pressure that averaged 0.5 percent higher than the auscultatory measurement. Diastolic pressure measurements averaged 2.5 percent lower than values determined by the auscultatory method. Figure 30 shows the "floating waveform" as obtained with the P23B transducer. The marker pulse represents one second of elapsed time.

C. A Single Transducer Digital Pickup

The small size and the increased pressure limit of the PG222 transducer designate it as being more suitable for use in a waveform determination system than the P23B transducer. A chief disadvantage to this use is the relative low output signal level as compared to the P23B transducer. Figure 31 shows the response of both transducers to the same waveform input. The top curve is for the PG222 transducer, and represents a signal level of approximately 900 μ v. The lower curve is the signal output of the P23B transducer and represents 2.6 mv. Excellent correspondence is noted between both transducer waveforms and published pulse waveforms (20).

The low level of the pulse waveform output signal from the PG222 transducer will be amplified by the Tektronix 122 "Low Level Preamplifier". A coupling capacitor of 1 μ f was selected to provide a time constant of one second when combined with the gird assistances of the amplifier. The top curve of Figure 32 shows the output of the PG222 transducer as an input to the preamplifier and the preamplifier output waveform of the preamplifier on the bottom curve. There exists a high degree of correlation between both waveforms as any distortion introduced by the preamplifiers is extremely small.

The use of the preamplifier in conjunction with the PG222 transducer provides a means of determining pulse waveforms and pressure information from the same transducer. Consideration was given to constructing a system as shown in block diagram form in Figure 6c. This construction is shown in Figure 28a. The model shown on the left of Figure 28a was con-

structed with 10 by 15 mm pickup and occlusion bags. The bags were connected internally by 17 mm of .58 mm inside diameter polyethylene tubing. The models were constructed using liquid latex. The transducer was attached to the top tube on the right side of the device. Provision for filling the device with fluid and for removing trapped air was incorporated in the middle tube. The bottom tube on the right was used to supply air pressure to the inflatable pressure chamber of the device. In practice the bag was placed against the digital arterial pulse area and bound with a nylon bandage. The fluid in the pickup and occlusion bags formed a closed system with the transducer diaphragm. As the pressure in the pressure chamber increased both the occlusion and pickup bags were forced against the arterial pickup area. Ideally, the mean pressure in the pickup system would be a measure of the applied pressure. When the pressure reached the value of systolic pressure the occlusion bag should have occluded the artery. There would be no pressure pulse transmitted through the pickup bag. The pressure pulse which is transmitted through the occlusion bag should be filtered. Only a mean pressure, corresponding to systolic pressure should be transmitted to the pickup bag and then to the transducer. It was found that the pressure of complete occlusion which corresponded to systolic pressure was 50 mm Hg higher than the value obtained by the auscultatory method. The filter system of this device failed to smooth the pulses transmitted through the occluding bag as described in a previous section. However, as the pressure within the occlusion bag increased above systolic pressure the tension on the bag surface also increased with a resultant decrease in the transmitting

ability of the effective occlusion diaphragm. Thus, the pulse waveform transmission decreased and an apparent systolic pressure was attained. The devices shown in Figure 28a failed to provide a reliable means of obtaining the blood pressure waveform.

Several filter devices were investigated. Two such devices are shown in Figure 28b. These filters consisted of a series combination of a .30 mm tube, an elastic expansion chamber, a .58 mm tube, and a second elastic chamber. This filter is equivalent to two stages of an RC filter. It was observed that a tube of small enough diameter to damp out the effects of pulse waveform variations with the associated small resultant flow of these variations would introduce considerable error in the transmission of the mean occlusion pressure with its relative high resultant fluid flow.

A system which is based upon the schematic diagrams of Figure 6c but which is free of the deterious effects of pulse transmission through the occlusion bag is shown in Figure 33. The pickup membrane connected to the transducer adapter and PG222 transducer is shown on the right of Figure 33. The occlusion bag is shown on the left. The middle bag serves to transmit the occlusion pressure from the occlusion bag to the pickup transducer system. In operation, the occlusion bag was placed around the proximal phalanx and wrapped by a nonstretchable nylon bandage. The pickup and transducer form a closed system filled with distilled water which was placed over the digital artery on the medial section of the middle phalanx. The second bag was wrapped around the pickup diaphragm and secured with nonstretchable nylon bandage. The two bags are connected by two .58 mm polyethelyene tubes connected in parallel to a 10 cc

syringe which is shown in the far left of Figure 33. A total of 130 cm of two parallel .58 mm inside diameter tubes connect the two bags.

As the pressure in the occlusion bag was varied, corresponding variations in pressure of the bag placed over the pickup diaphragm resulted. The variation of pressure in the bag around the pickup diaphragm was transmitted to the transducer as a mean pressure change. Pulse variation due to pulse pressure on the occlusion bag was effectively filtered through the use of an air transmitting medium and the long connecting tubes. The pressure in the occlusion bag subsystem was controlled by changing the position of the plunger of the syringe.

Several tests were conducted using the system described above. Figure 34 is a representation of the waveforms obtained with this system. Figure 35 shows the system applied to the index finger of a test subject. Figure 34a shows the output of the PG222 transducer after amplification by the preamplifier. This curve is displayed through a capacitive coupled amplifier and permits the waveform to be displayed without the effects of changes in mean pressure. The direct transducer output is shown in Figure 34b. The occlusion pressure was increased to 140 mm Hg at which time the pulse waveform output was zero. As the occlusion pressure was decreased to the level of systolic pressure the pulse waveform remained zero. At 105 mm Hg, which corresponds to systolic pressure, the curve of Figure 34b shows an initial pulse waveform.

Criterion for the measurement of diastolic pressure has not been completely established. The use of a criterion based on the maximum pulse amplitude is not applicable as shown by Figure 34b. The amplitude of the

pulse waveform is dependent on the arterial pressure pulse in addition to the pressure of the occlusion subsystem. Diastolic pressure of the subject used to obtain Figure 34 was 70 mm Hg as measured by the auscultatory method. An approximate determination of diastolic blood pressure was made using the criterion that diastolic pressure was attained when the pulse waveform showed no further waveform distortion with release of the occlusion pressure. This criterion requires comparison of subsequent pulse waveforms. Due to the small value of the slope of the blood pressure waveform near the point at which diastolic pressure is attained, the actual comparison of waveform is relatively inaccurate and deviations in pressure reading of 20 percent from those obtained by the auscultatory method have been obtained. The comparison of the pressure waveforms can be made by taking the derivative of the section of the pressure waveform from the dicrotic notch to the point of diastolic pressure. The instant at which this derivative attains a constant value corresponds to diastolic pressure. The circuitry to perform this operation has not been constructed.

As shown in Figure 35, the combined pickup, occlusion, and transduction device is small. The output signal is essentially free from sperious signals due to muscle movement. An even smaller, more compact device is possible if the transducer is attached directly to the pickup diaphragm without using the adapter. The location of the syringe or pressure source is not critical and an extension tube could be connected between the syringe and the tee connector of the occlusion bags.

D. Observations of Experimental Results

Several interesting results were observed in compiling the data presented above. Ten subjects were used in the above tests. Several experimental determinations were made on each subject. In all cases it was observed that the initial blood pressure readings were higher than those readings taken after the subject became accustomed to the equipment and procedures. Systolic blood pressure readings were observed to vary greatly if the subject was permitted to become exposed to situations that produced tension or anxiety. On several occasions the systolic pressure readings were found to raise as much as 10 percent after the injection of a provoking or embarrassing question to the subject. This result suggests the use of above apparatus as a "lie detection" device to replace the auscultatory device described by Marston (9).

It was observed that if the occlusion pressure was maintained in excess of the systolic pressure a decay occurred in the mean level of pickup representation tracings taken with a pickup diaphragm placed on the finger and an occlusion cuff placed on the arm. If the occlusion pressure is maintained for a period of 10 seconds the decay will attain an essentially constant level. The shape of the decay curve suggests an exponential type decay similar to the discharge of a capacitor. It has been found that as the occlusion pressure is released to below systolic pressure the mean pressure level of the pulse representation will increase to a maximum at diastolic pressure, which is in excess of the normal level before the application of the occlusion pressure. The mean pressure level will return to the normal pressure level as the occlusion pressure is reduced to zero.

In waveforms obtained in finger occlusion tests a positive variation of the mean pressure level has been observed if the occlusion pressure is maintained in excess of the systolic pressure. The variation in the pulse representation appears similar to the charging of a capacitor. As the occlusion pressure is decreased, the waveform returns to the same mean level as attained prior to occlusion.

Figure 36 shows the result of a simultaneous representation of an Electrocardiogram and blood pressure waveform. The apparent velocity of waveform propagation may be obtained using an approximate vessel length of 1 m between the digital arterial pulse pickup area and the left ventricle of the heart, and a time lag of .14 seconds between the polarization waveforms of the ventricle, as obtained from Figure 36. The apparent velocity of propagation can be calculated as 7.15 mps as compared with the published value of 9 meters per second (2).

The variation in mean pulse pressure with respiration changes is shown in Figure 37. At time A a deep inspiration was taken. At time B the inspiration was released and normal breathing resumed. The mean pressure corresponds approximately to the level of the dicrotic notch as shown in Figure 37. The dicrotic notch appears as a broad line through the pulse waveform of Figure 40. The time scale of this figure is five seconds per division. Variations in the heart rate are obtainable by observing the change in the period of the pulse waveform. Clynes (5) has examined the factors concerned with respiratory control of the heart rate. Figure 40 appears to be a representation of the effects of respiration.

It was observed that the value of pressure corresponding to the ap-

pearance of the dicrotic notch remained essentially constant for each subject while the value of systolic pressure varied over the test period.

VI. CONCLUSION

An investigation of the classical methods of Sphygmomanometry has been made. The circulatory system has been reviewed to determine the anatomical and hemodynamical considerations of blood pressure waveform determination. A system for obtaining the "floating representation" of the pressure pulse has been designed, constructed, and evaluated. Two systems for obtaining the blood pressure waveform by extra-arterial means have been designed, constructed, and tested.

A waveform representation having excellent correspondence to published waveforms (20) has been obtained. Devices for obtaining the blood pressure waveform including the reference values of systolic and diastolic pressures have been constructed. These devices are small, easily applied, and essentially free of the effect of body movement. The systems developed have been used in cyclic pressure determinations for periods of 90 minutes without deterious effects.

One system of waveform determination requires the use of two pressure transducers. With this system both systolic and diastolic pressures can be determined and correlated to the "floating representation". The second system is extremely compact, utilizes one pressure transducer, and provides an accurate method for determining systolic pressure. Circuitry in addition to a recording system will be required for the satisfactory determination of diastolic pressure.

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Both members of the Biomedical Electronics Group and the United States Air Force Institute of Technology Group volunteered to be test subjects. The author is indebted to the United States Air Force for making this period of Graduate Study possible. The author wishes to express his gratitude to his wife, Nancy A Gowen, for her unfailing encouragement and for her assistance in editing the original manuscript.

IX. APPENDIX

A measure of the mean pressure within an elastic vessel can be obtained by applying an oscillating mass to the external vessel walls and observing the resultant frequency of oscillation. Figure 38 shows an elastic tube of radius a. The dashed line tube is the resultant form of the elastic tube when the vessel wall is deformed a radial distance by two parallel planes. This deformation would correspond to forcing a plane surface against vessel over an axial length ℓ and for a radial distance γ, or a reduction in the original diameter in the z direction of 2γ. The force to cause the deflection γ may be written as

$$F = P\ell A \tag{1}$$

where P is the pressure inside the vessel and ℓ is the length over which the force is applied. Assuming that the undeformed section of the vessel maintains the original circular curvature, the area of deformation may be written as

$$A = 2a\left(\frac{\pi}{2} - \theta\right) \tag{2}$$

where a is the radius and θ is the angle measured as shown in Figure 38. F may be written as

$$F = P\ell 2a\left(\frac{\pi}{2} - \theta\right) \tag{3}$$

Utilizing the trigonometry relationships of the figure, equation 3 may be

written

$$4 \qquad F = P\ell 2a\left(\frac{\pi}{2} - \sin^{-1}\left(1 - \frac{\gamma}{a}\right)\right)$$

An effective spring constant k may be obtained by writing $k = \frac{dF}{d\gamma}$. Differentiating equation 4 with respect to γ

$$5 \qquad K = \frac{2P\ell}{\sqrt{1 - (1 - \frac{\gamma}{a})^2}}$$

The undamped natural frequency for an oscillating spring-mass system is given (1) as

$$6 \qquad f_n = \sqrt{\frac{K}{M}}$$

where M is the mass. The undamped natural frequency may be expressed in terms of the pressure within the elastic vessel by equation 5

$$7 \qquad f_n = \sqrt{\frac{2P\ell}{M\sqrt{1-(1-\frac{\gamma}{a})^2}}}$$

Figure 39 is a plot of the variation of $F/2P\ell a$ as a function of γ/a. An essentially linear portion of the resultant curve occurs about the value of $\gamma/a = .5$.

Assuming a pressure of .142 atmospheric pressure, which corresponds to 108 mm Hg, an effective deformation length of 1 cm and a 5 gm mass, the undamped natural frequency for a deformation $\gamma = a/2$ can be calculated as 42 cps.

X. FIGURES

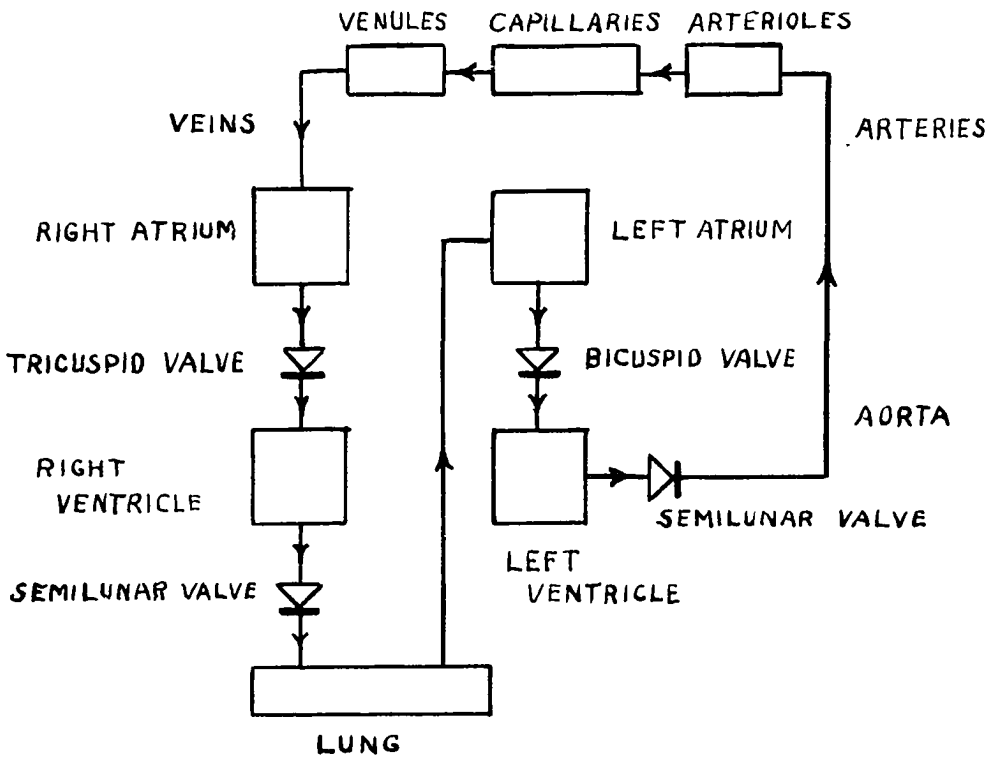


FIGURE 1. THE CIRCULATORY SYSTEM

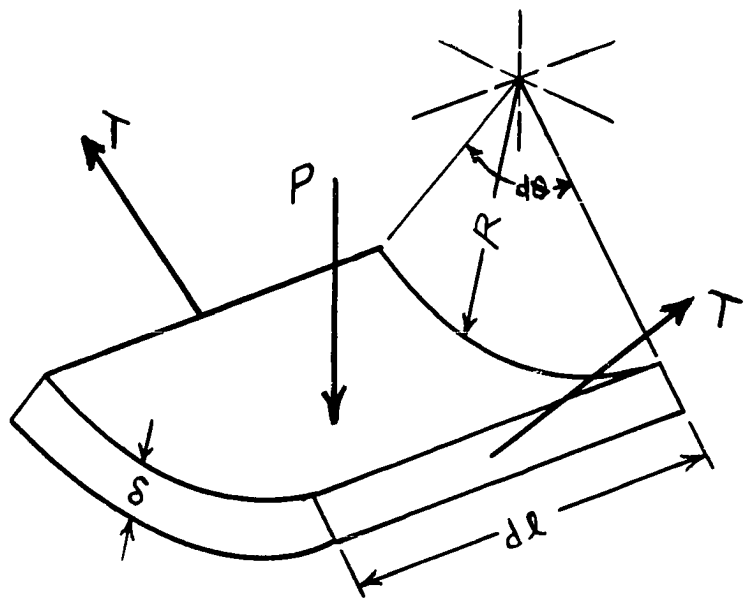


FIGURE 2. HEMODYNAMIC FORCE MODEL

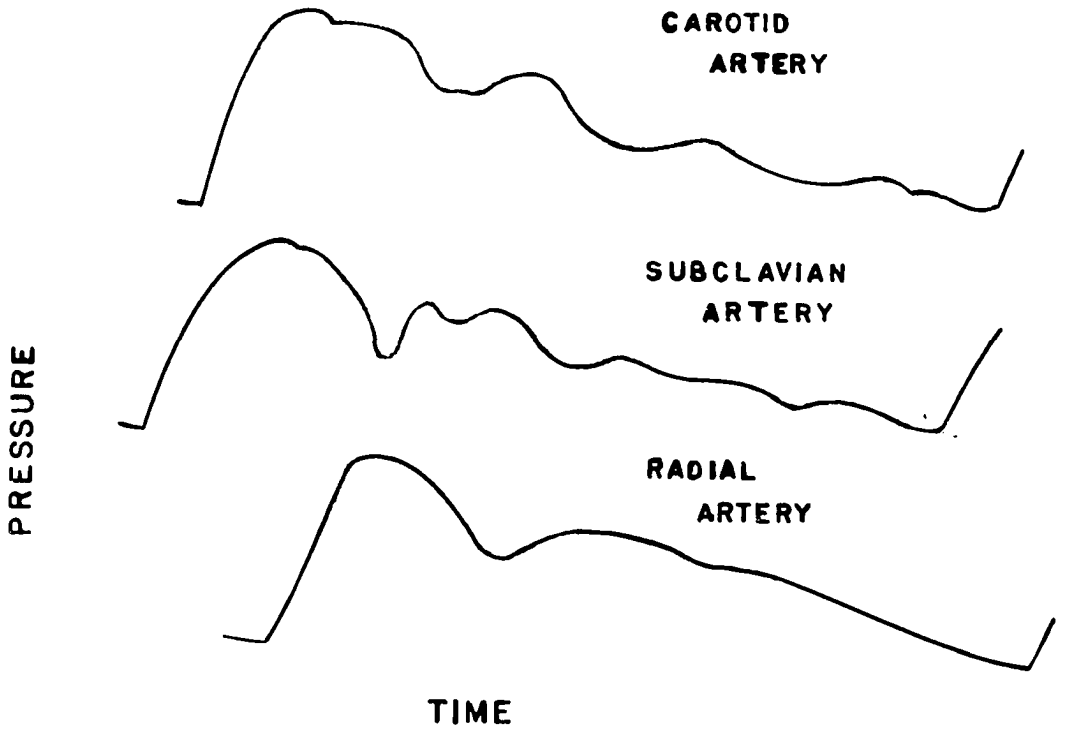


FIGURE 3. BLOOD PRESSURE WAVEFORMS
(ADAPTED FROM WIGGERS (20))

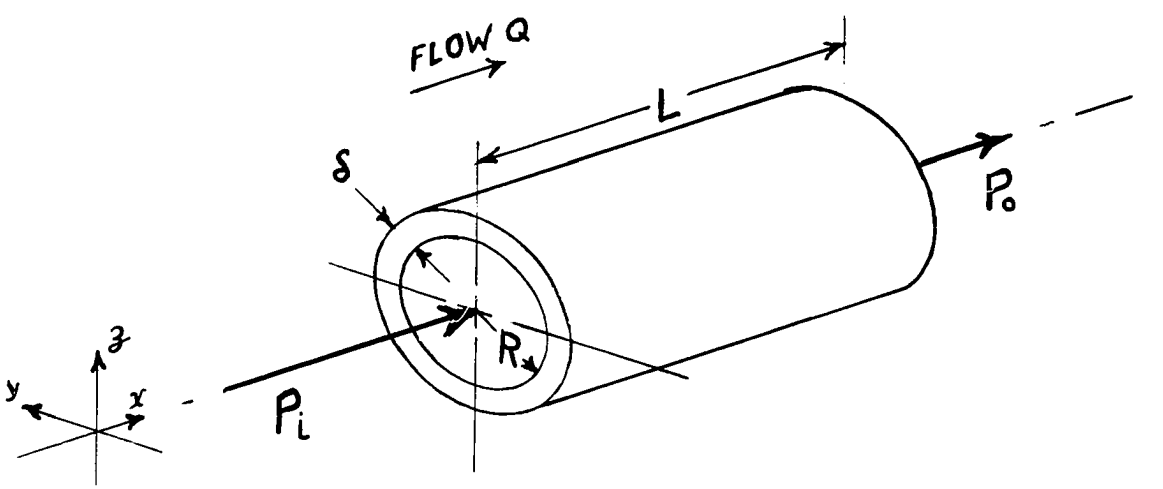


FIGURE 4. DIAGRAM OF AN ELASTIC VESSEL

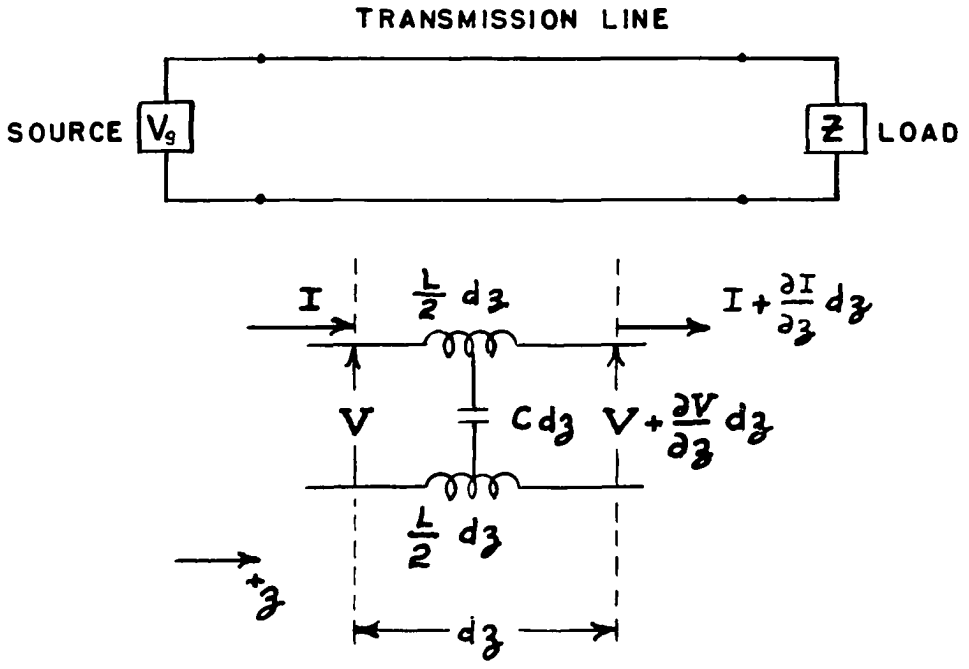


FIGURE 5a. TRANSMISSION LINE

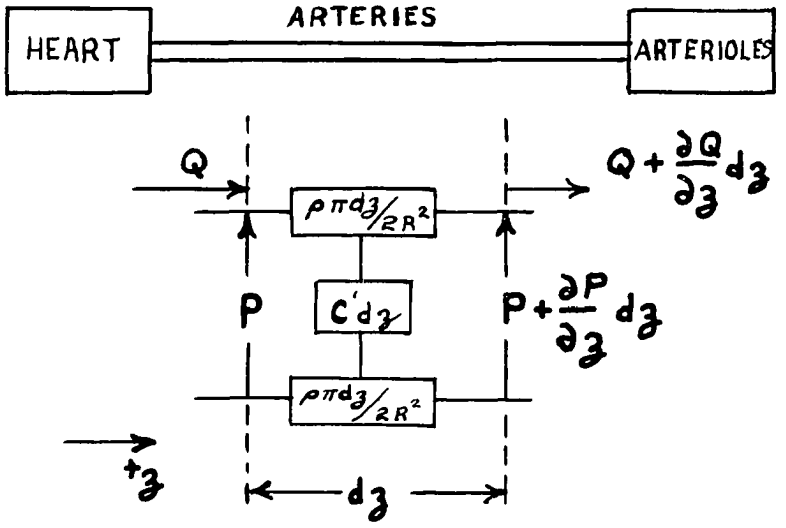


FIGURE 5b. TRANSMISSION LINE ANALOG OF ARTERIAL SYSTEM

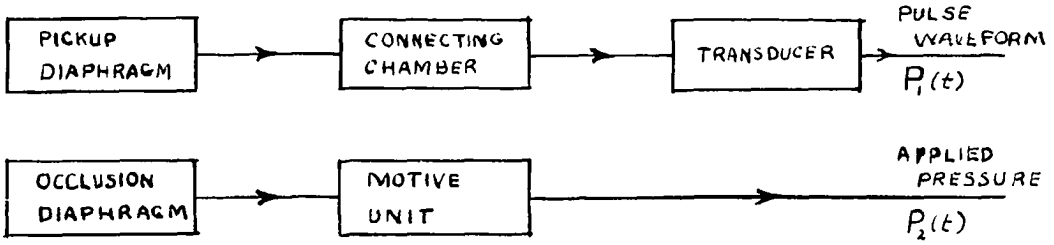


FIGURE 6a. TRANSDUCER - PRESSURE DEVICE

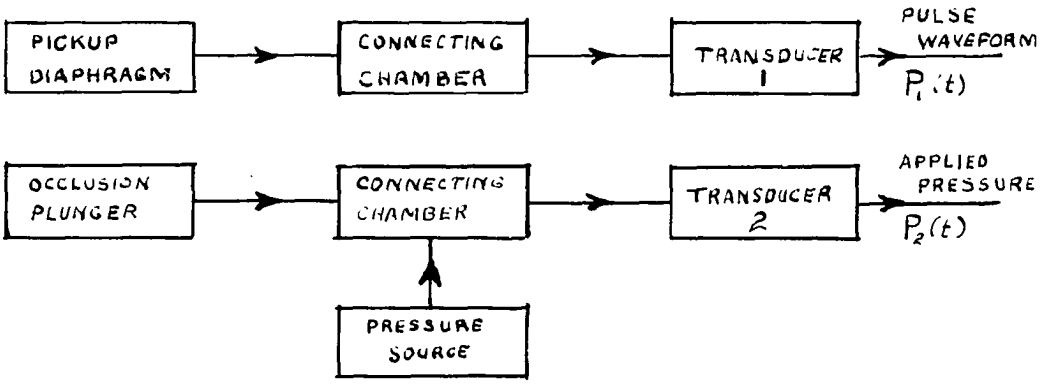


FIGURE 6b. TWO TRANSDUCER DEVICE

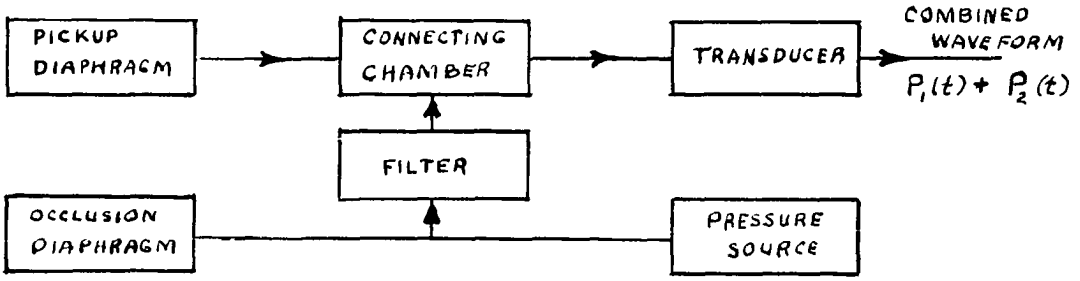


FIGURE 6c. SINGLE TRANSDUCER DEVICE

FIGURE 6. DEVICES FOR OBTAINING THE BLOOD PRESSURE WAVEFORM

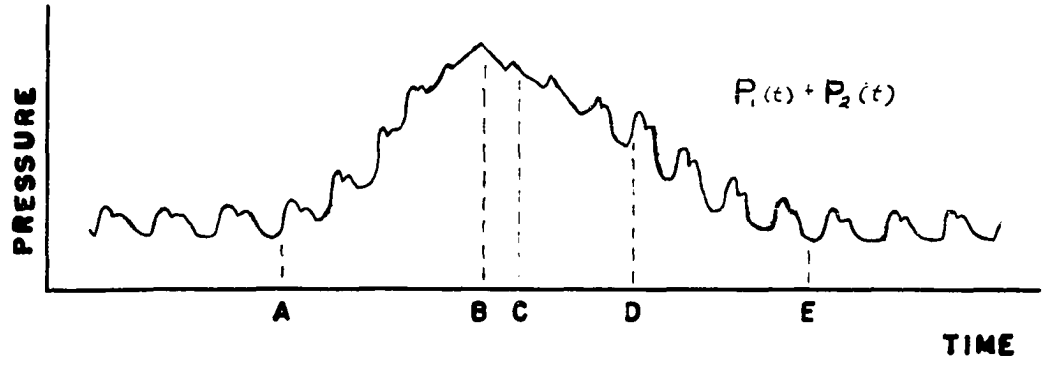


FIGURE 7. REPRESENTATIVE OUTPUT OF THE SINGLE TRANSDUCER DEVICE

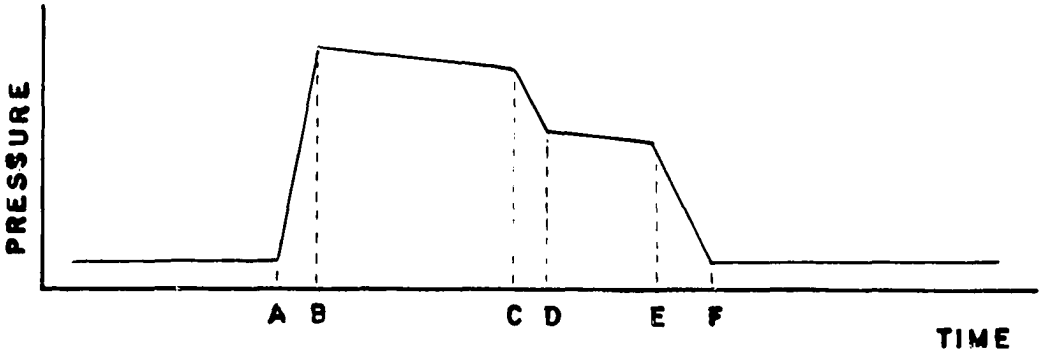


FIGURE 8. IDEAL LOCUS OF APPLIED PRESSURE

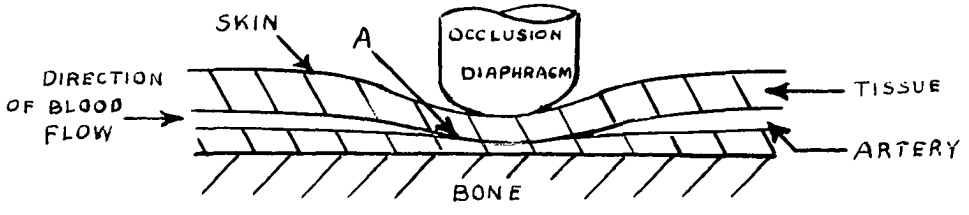


FIGURE 9. ARTERIAL OCCLUSION

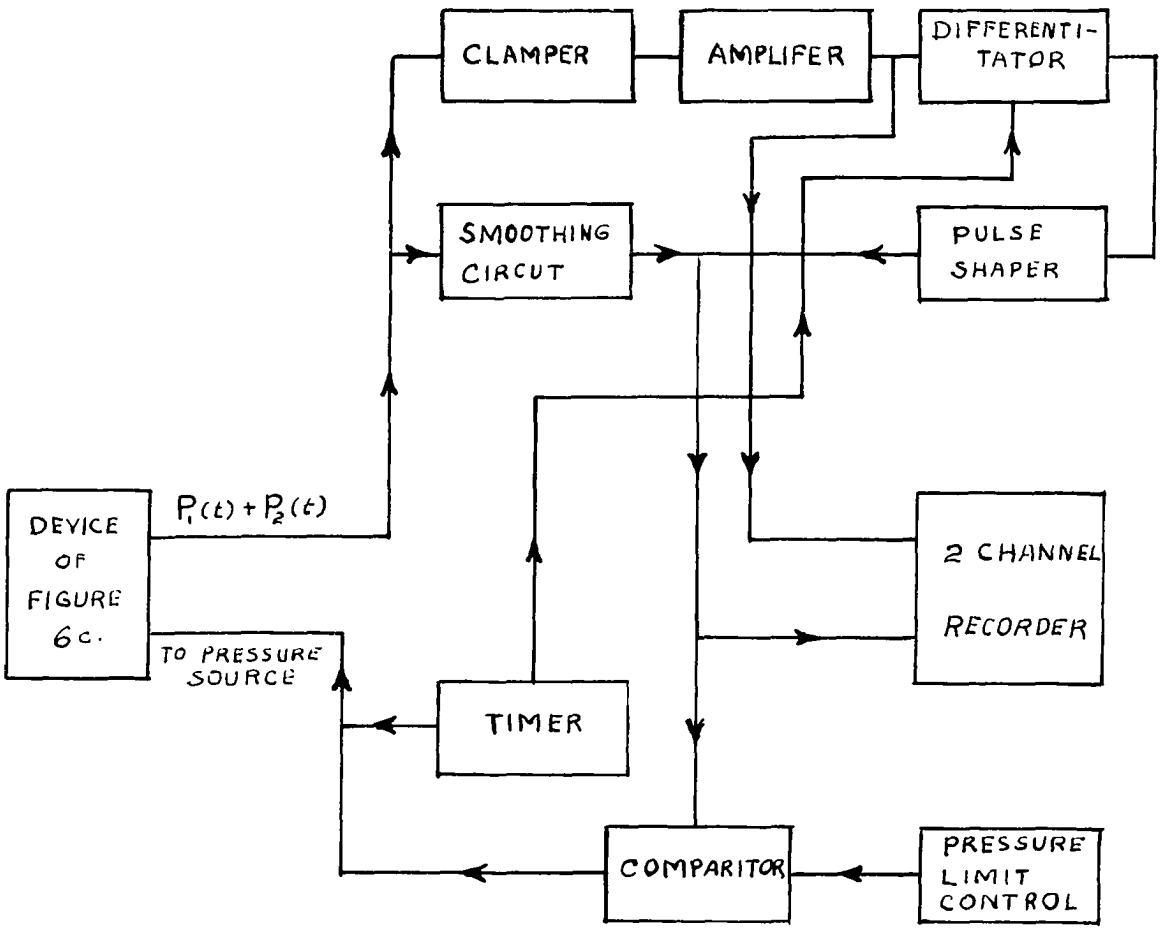


FIGURE 10. SYSTEM FOR OBTAINING THE BLOOD PRESSURE WAVEFORM

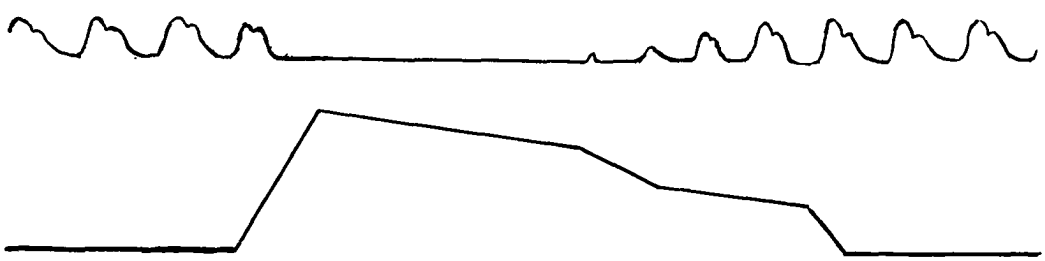


FIGURE 11. RECORDED OUTPUT

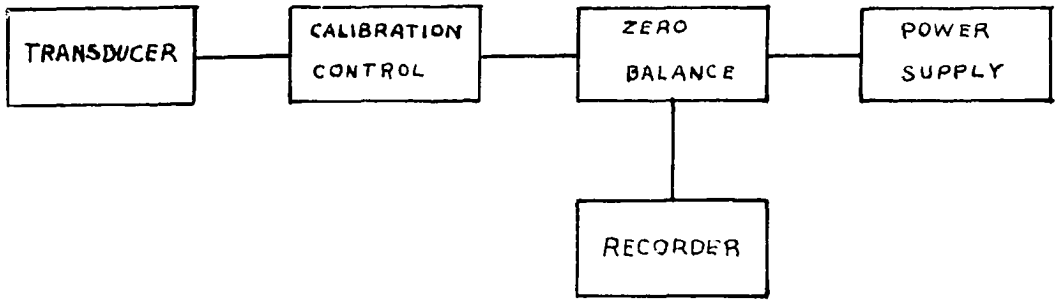


FIGURE 12. TRANSDUCER POWER-BALANCE SYSTEM

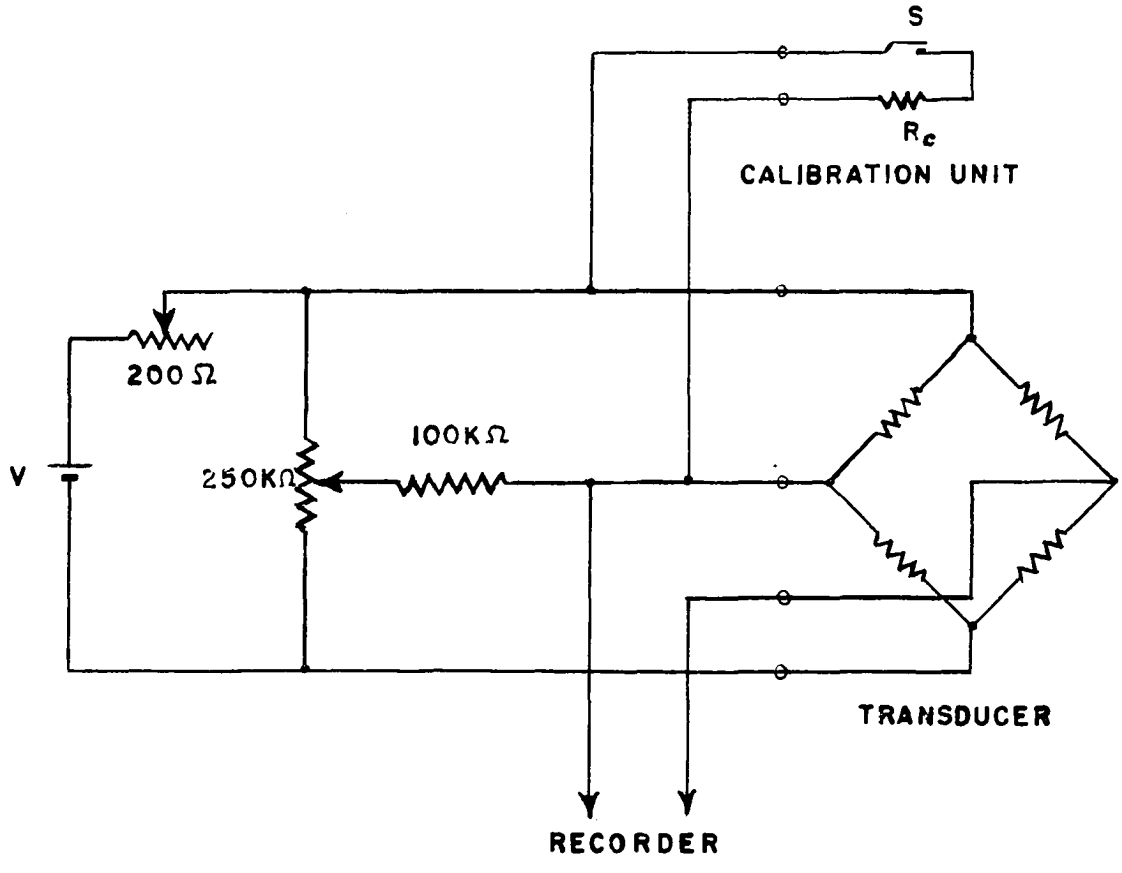
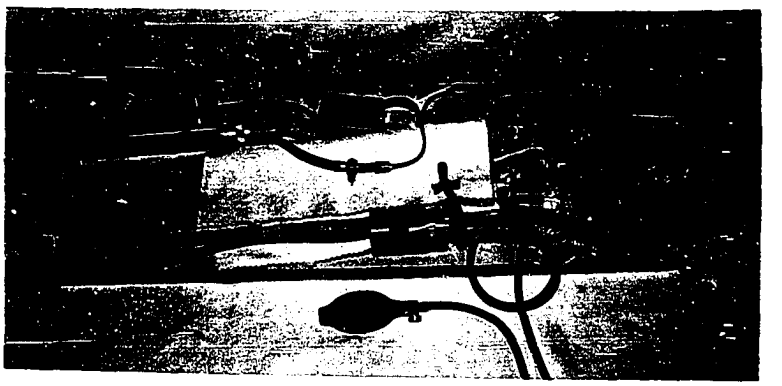


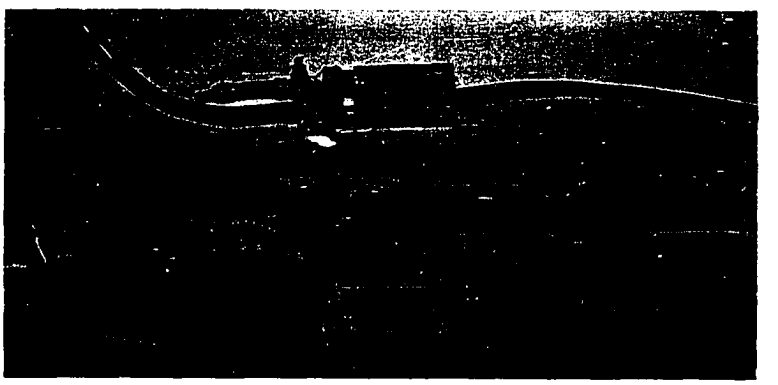
FIGURE 13. POWER-BALANCE UNIT

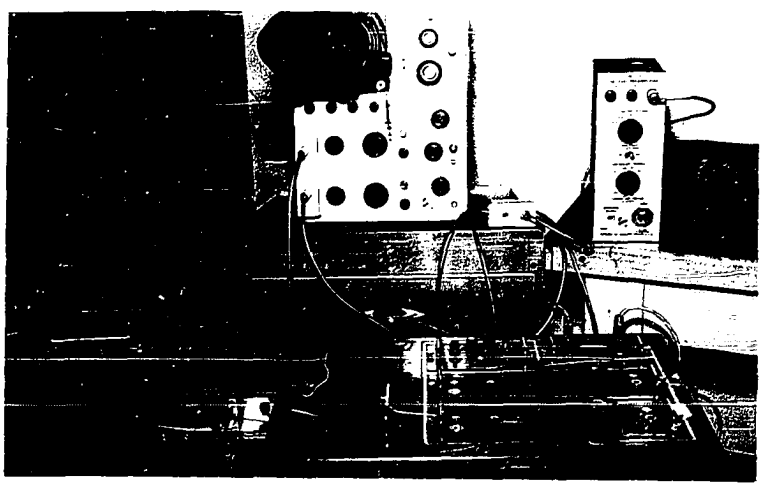
Figure 14. Test assembly

Figure 15. P23B and PG222 transducers

Figure 16. Recording equipment and Tektronix preamplifier







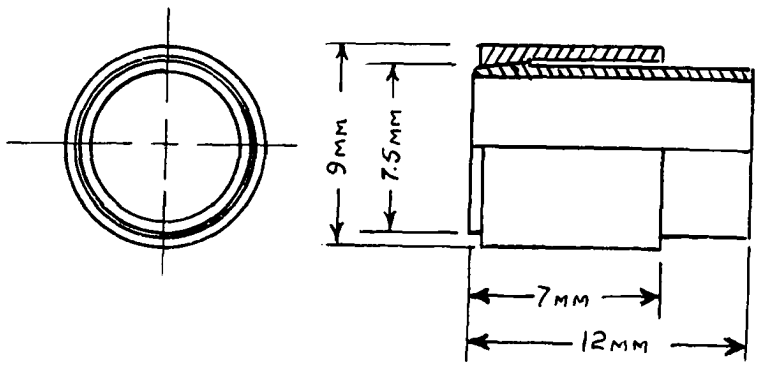


FIGURE 17a. DIAPHRAGM HOLDER

FIGURE 17b. SPHERICAL DIAPHRAGM

FIGURE 17c. TAUT DIAPHRAGM

FIGURE 17d. BUTTON DIAPHRAGM

FIGURE 17. TEST DIAPHRAGMS

Figure 18. Temporal test platform

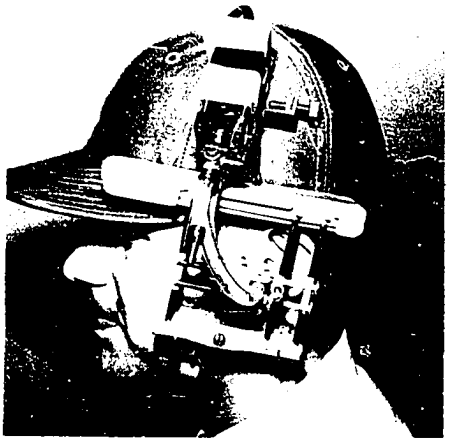
Figure 19. Temporal arterial pulse area waveforms

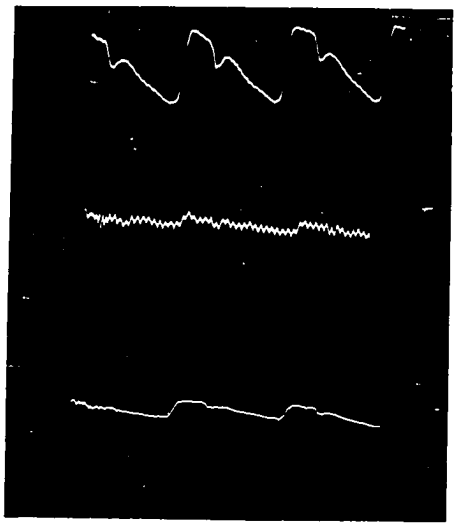
Figure 20a. Digital pickup diaphragm a

Figure 20b. Digital pickup diaphragm b

Figure 20c. Digital pickup diaphragm c

Figure 20d. Digital pickup diaphragm d





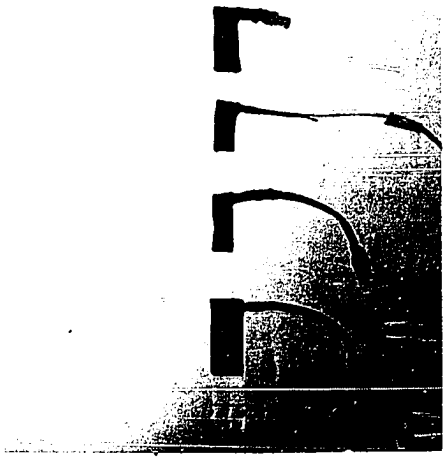
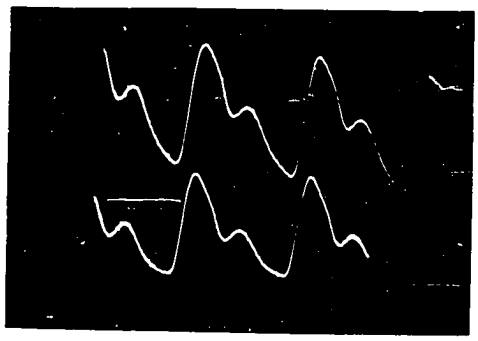


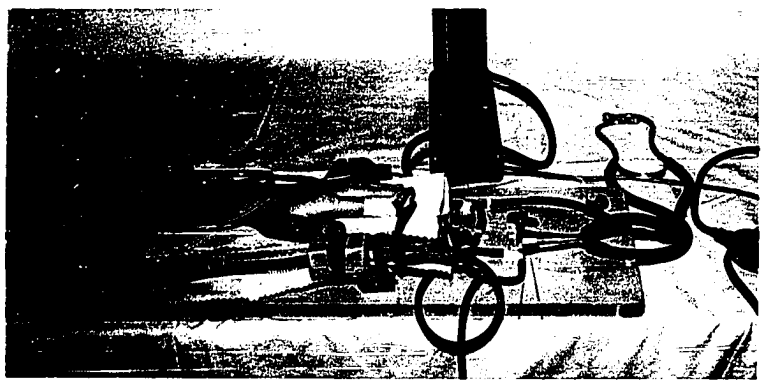
Figure 21. Digital pulse area "floating" waveform representation

Figure 22. Test assembly for obtaining the digital "floating" waveform representation

Figure 23a. Temporal "floating" waveform representation

Figure 23b. Digital "floating" waveform representation





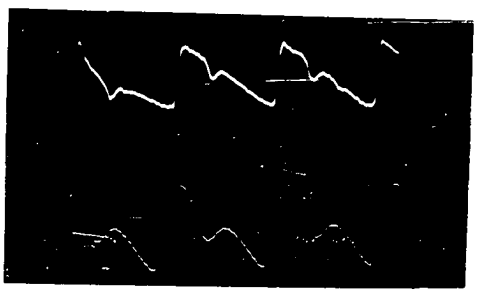
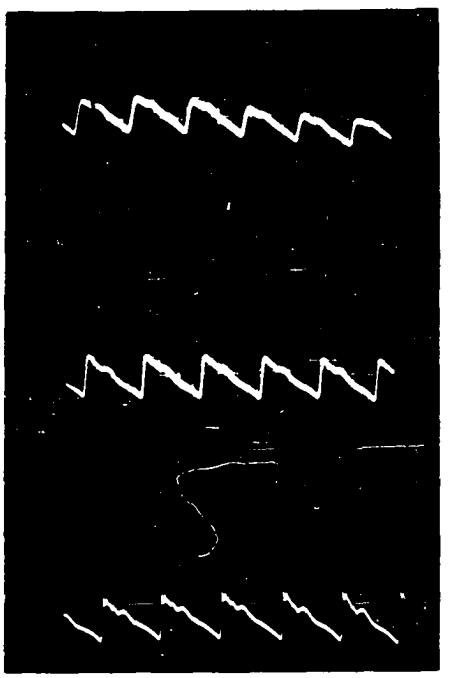


Figure 24a. Waveform obtained with digital pickup A

Figure 24b. Waveform obtained with digital pickup B

Figure 24c. Waveform obtained with digital pickup C

Figure 25. Digital occlusion and pickup diaphragms



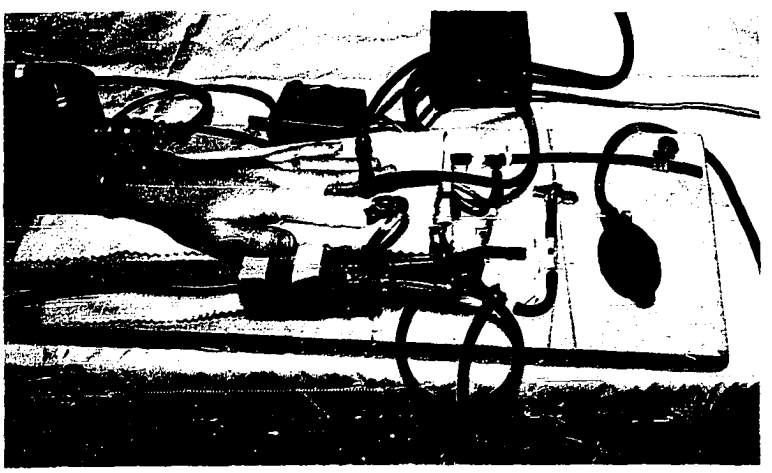


Figure 26. Blood pressure waveform determined with a brachial occlusion cuff and a digital pickup diaphragm

Figure 27. "Floating" blood pressure waveform showing the effect of respiration

150
100
80
0
mm Hg

Figure 28a. Single transducer devices

Figure 28b. Filter devices

Figure 28c. Occlusion diaphragms

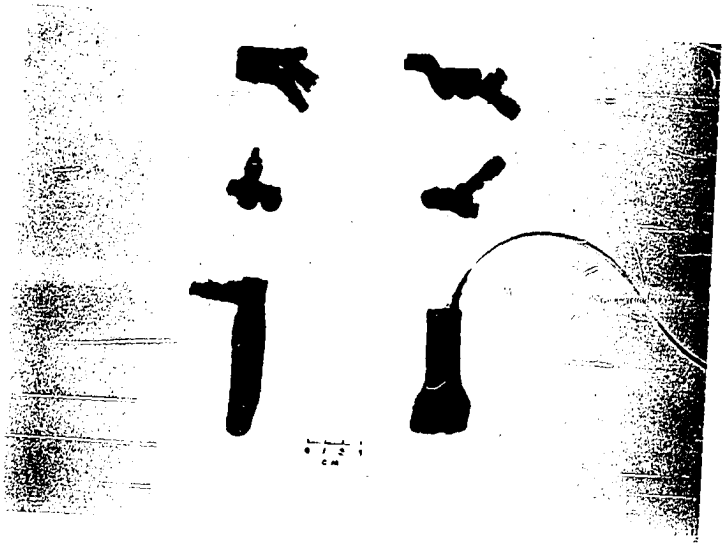


Figure 29. Blood pressure waveform determined with digital occlusion and pickup diaphragms

Figure 30. Digital "floating" waveform representation as recorded by the Sanborn Recorder

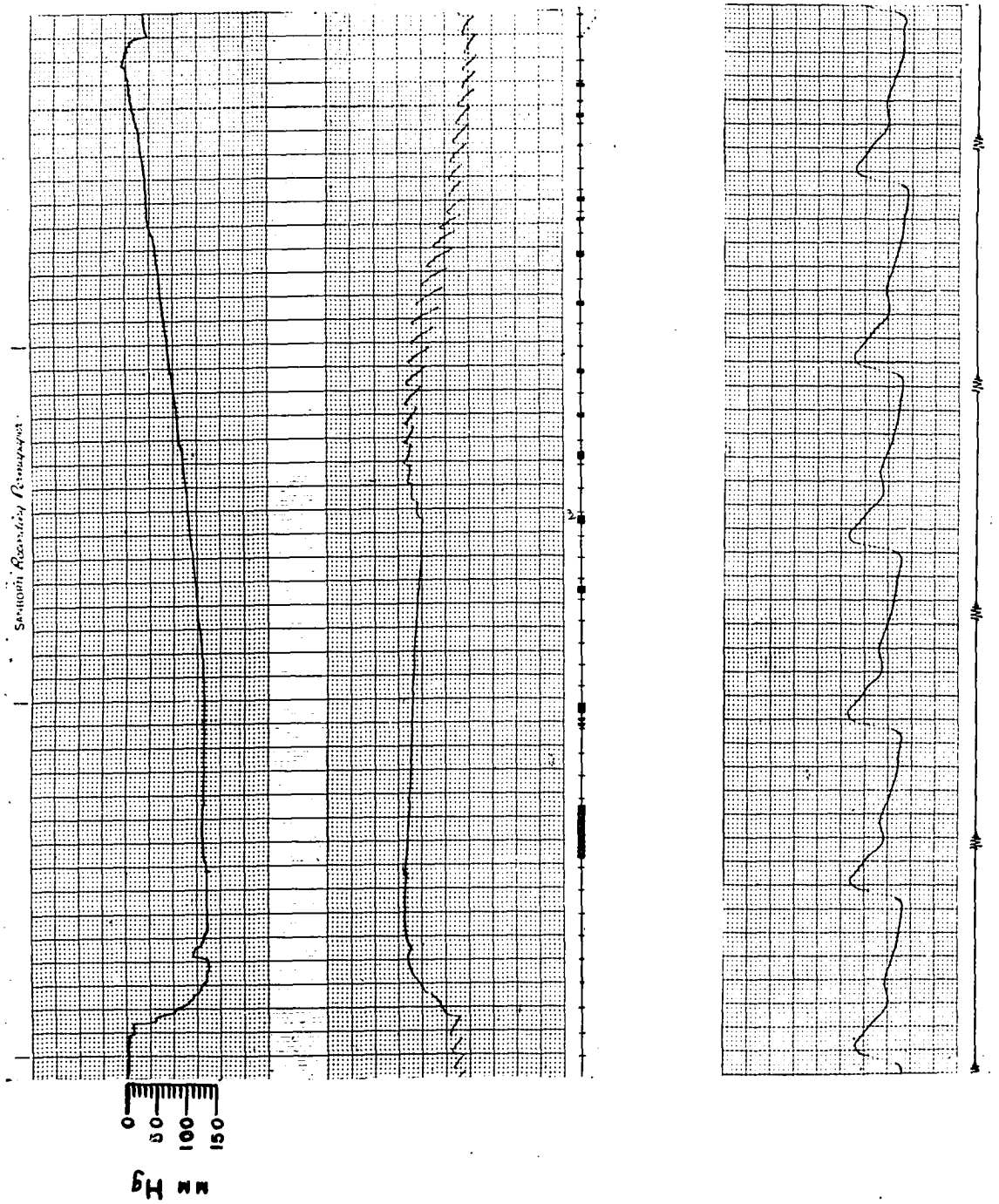
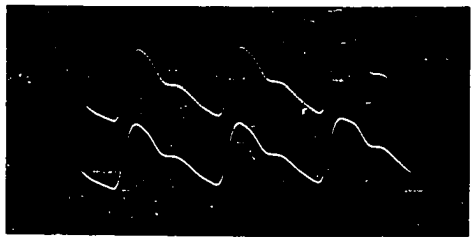
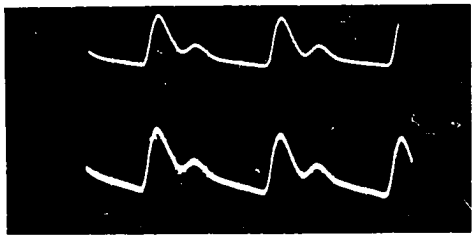

Самопи Ковалёва Ленинград
0
50
100
150
мм Hg

Figure 31. Comparison of the output waveform representation of the P23B and PG222 transducers when subjected to the same input waveform

Figure 32. Comparison of input and output waveforms of the Tektronix 122 preamplifier

Figure 33. Single transducer device for obtaining the blood pressure waveform





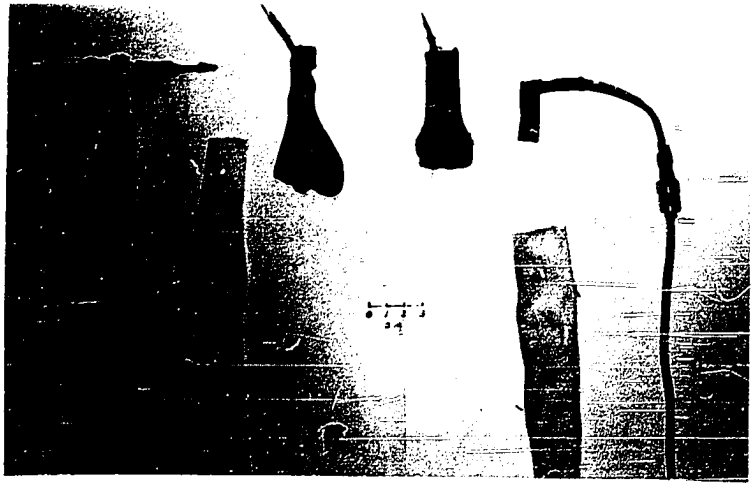
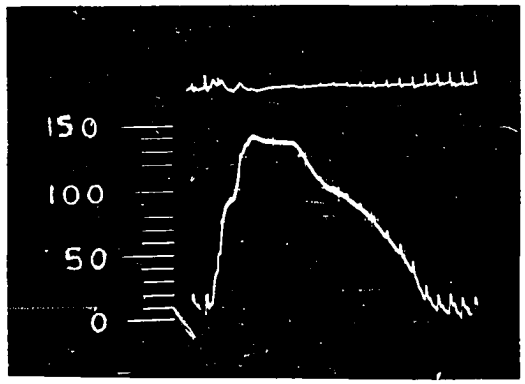


Figure 34. Blood pressure waveform obtained with the single transducer device

Figure 35. Single transducer device applied to the index digit of subject


150
100
50
0

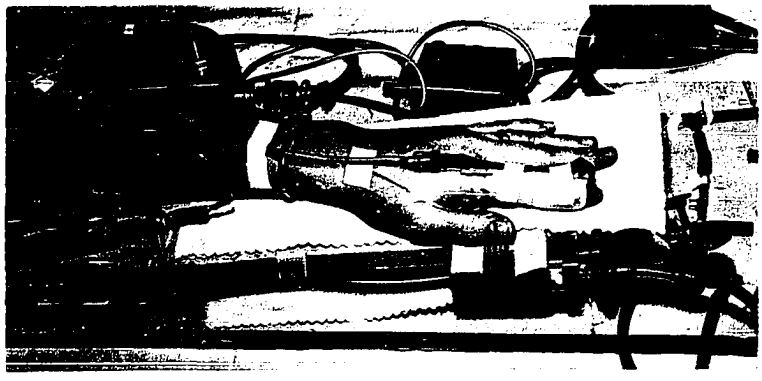
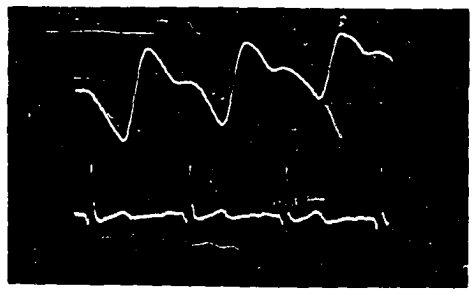
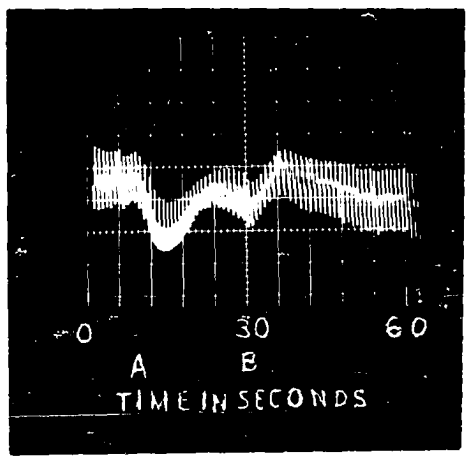


Figure 36. Simultaneous recording of electrocardiogram and pressure waveform

Figure 37. "Floating" waveform representation showing the effect of respiratory changes




0
30
60
A
B
TIME IN SECONDS

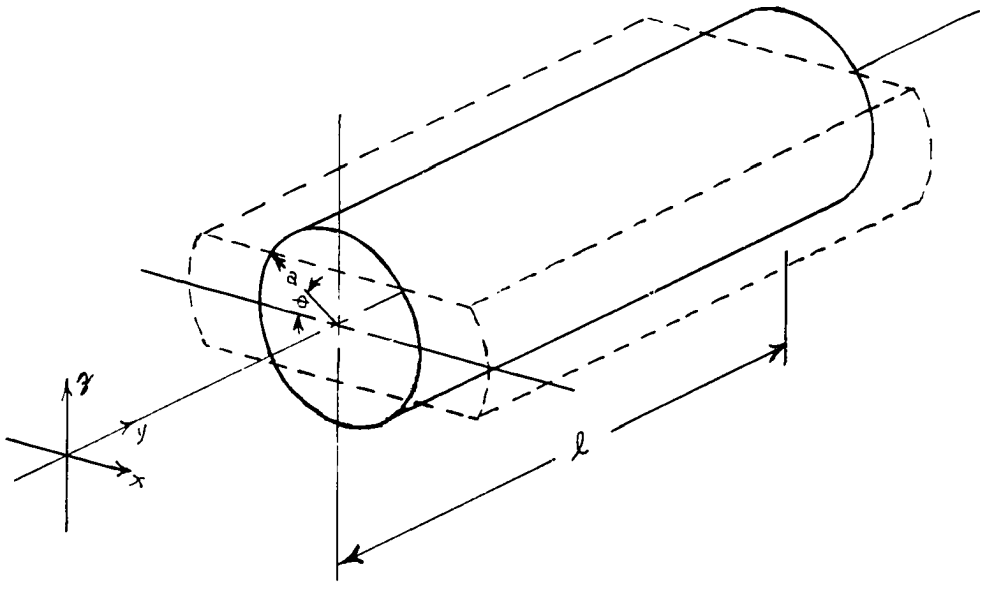


FIGURE 38. MODEL OF AN ELASTIC TUBE WITH NEGLIGIBLE WALL THICKNESS

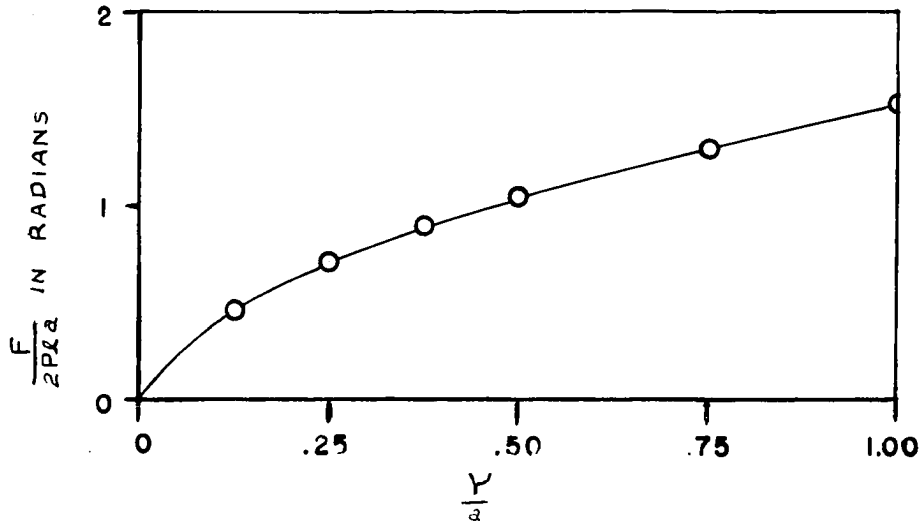


FIGURE 39. GRAPH OF THE VARIATION OF $\frac{F}{2Pla}$ AS A FUNCTION OF $\frac{r}{a}$

XI. TABLES

Table 1. Quantities for an arterial system analog using a loss-less transmission line

Arterial System		Transmission Line	
Quantity	Symbol	Quantity	Symbol
Pressure	P	Voltage	V
Flow	Q	Current	I
Volume	V	Charge	q
Capacitance factor	C'	Capacitance	C
Fluid factor	ρ / R_o^2	Inductance	L

Table 2. Characteristics of pressure transducers used in the devices described

Quantity	Units	P23B Transducer	PG222 Transducer
Dimension	cm	7.6 x 2.8 x 2.8	1.42 x .78(Dia)
Weight	gm	453	3
Pressure range	mm Hg	0 to 50	0 to 516
Input resistance, R_{in}	ohms	334.0	198.2
Output resistance, R_o	ohms	334.0	198.1
Calibration factor, F	μv/mm Hg	360.0	40.2
Excitation, E	volts	12	3
Calibration resistor, R_c	ohms	11,310 for 20 mm Hg	36,900 for 100 mm Hg
Temperature correction	^oF	not corrected	-65 to 250
Approximate natural frequency	cps	1000	7000

Table 3. Temporal arterial pulse area evaluation

Waveform of Figure 23	Diaphragm Shape of Figure 17	Maximum Amplitude mv	mm Hg	Waveshape Evaluation
Top	Spherical	2.2	6.1	Good
Center	Taut	0.1	0.28	Poor
Bottom	Button	0.6	1.7	Fair

Table 4. Comparison of auscultatory and waveform methods for obtaining systolic and diastolic pressure

	Pressures by the Auscultatory Method		Pressures by the Waveform Method			Percentage Difference	
Test Subject	Systolic in mm Hg	Diastolic in mm Hg	Systolic in mm Hg	Diastolic in mm Hg	Dicrotic Notch in mm Hg	Systolic	Diastolic
1	105	65	109	60	70	3.8	-8.2
2	121	59	119	58	60	-2.5	1.7
3	108	72	111	60	88	2.8	-16.7
4	118	70	118	65	75	0	-7.1
5	112	70	105	60	60	-5.8	-14.2
6	101	70	95	70	70	-5.9	0
7	110	78	110	75	90	0	-3.8
8	100	78	100	80	85	0	2.6
9	101	70	98	60	75	-3	-14.2
10	118	70	115	70	90	-2.5	0
Average						-1.3	-5.9

Table 5. Comparison of systolic and diastolic pressure determined by arm and digit occlusion cuffs

	Pressures with 130 mm Arm Cuff		Pressures with 25 mm Digit Cuff			Percentage Difference	
Test Subject	Systolic in mm Hg	Diastolic in mm Hg	Systolic in mm Hg	Diastolic in mm Hg	Dicrotic Notch in mm Hg	Systolic	Diastolic
1	103	70	105	65	90	1.9	-7.1
2	110	78	110	75	90	0	-3.8
3	118	72	120	70	95	1.1	-2.9
4	118	70	117	68	75	-0.8	-2.9
5	98	60	98	62	70	0	3.3
Average						0.5	-2.5